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Intrinsic foot muscles act to stabilise the foot when greater fluctuations in centre of pressure movement result from increased postural balance challenge

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ABSTRACT

Background: Increased postural balance challenge is associated with more fluctuations in centre of pressure movement, indicating increased interference from the postural control system. The role of intrinsic foot muscles in balance control is relatively understudied and whether such control system interference occurs at the level of these muscles is unknown.
Research Question: Do fewer fluctuations in intrinsic foot muscle excitation occur in response to increased postural balance challenge?
Methods: Surface EMGs were recorded using a grid of 13 × 5 channels from the plantar surface of the foot of 17 participants, who completed three balance tasks: bipedal stance; single leg stance and bipedal tip-toe. Centre of pressure (CoP) movement was calculated from simultaneously recorded force plate signals. Fluctuations in CoP and EMGs for each task were quantified using a sample entropy based metric, Entropy Halflife (EnHL). Longer EnHL indicates fewer signal fluctuations.
Results: The shortest EMG EnHL, 9.27 ± 3.34 ms (median ± interquartile range), occurred during bipedal stance and the longest during bipedal tip-toe 15.46 ± 11.16 ms, with 18.80 ± 8.00 ms recorded for single leg stance. Differences were statistically significant between bipedal stance and both bipedal tip-toe ($p < 0.001$) and single leg stance ($p < 0.001$). CoP EnHL for both anterior-posterior and medial-lateral movements also differed significantly between tasks ($p < 0.001$, both cases). However, anterior-posterior CoP EnHL was longest for bipedal stance 259.84 ± 230.22 ms and shortest for bipedal tip-toe 146.25 ± 73.35 ms. Medial-lateral CoP EnHL was also longest during bipedal stance 215.73 ± 187.58 ms, but shortest for single leg stance 113.48 ± 83.01 ms.
Significance: Fewer fluctuations in intrinsic foot muscle excitation occur in response to increased postural balance challenge. Fluctuations in CoP movement during balance must be predominantly driven by excitation of muscles extrinsic to the foot. Intrinsic foot muscles therefore likely play a greater role in stabilisation of the foot than balance control during the postural tasks studied.

1. Introduction

The human foot is critical for a wide range of activities of daily living. However, there are many pathologies that have direct clinical implications for foot function, influencing factors such as balance and mobility [1] and negatively affecting quality of life [2]. The complex anatomy of the human foot includes four layers of intrinsic foot muscles in the plantar region. Their location provides challenges to quantifying behaviour and function during weight bearing tasks. Recently however, it has been shown it is possible to record myoelectric signals (EMG) from these plantar foot muscles using flexible, multi-channel surface electrode arrays [3]. This provides new opportunities to investigate the behaviour of intrinsic foot muscles in postural balance control.

It is widely understood that maintenance of standing balance is dynamic, where the term dynamic indicates evolving or changing over time [4]. The neuromuscular system may solve the postural balance task by selecting from an array of available solutions in the motor system, all which provide the same result [5]. More challenging tasks likely constrain the number, or suitability, of the motor solutions available. One way to study the motor control of such time evolving

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tasks is by applying entropy-based measures, such as Sample Entropy (SampEn) [6]. SampEn measures the regularity within a signal by identifying instances of similarity. Recently a new metric, based on SampEn analyses, Entropy Halflife (EnHL) has been proposed [7]. It quantifies the timescale over which short-term fluctuations in a signal persist, with larger values indicating fewer fluctuations in the analysed signal. Longer EnHL may therefore indicate fewer solutions were available to, or used by, the system to solve the task [8,9] or that the system made fewer adjustments whilst completing the task [9].

EnHL has previously been applied to both EMG [8,10–12] and other signals related to postural mechanics such as centre of pressure (CoP) [9,13]. During bipedal standing, CoP EnHL is significantly longer than during single leg standing [9,13], indicating more short-term fluctuations in CoP occur during the more challenging balance task (single leg standing), particularly in the medial-lateral direction [9]. This has been suggested to indicate that the postural control system responds to the increased balance challenge of single leg standing by: i) increasing the frequency of adjustments to CoP motion; and/or ii) tolerating the use of a wider range of mechanically equivalent solutions (i.e. different configurations, with similar energetic characteristics) [9]. Whether such control system responses are reflected in altered regulation at the level of the skeletal muscles, and particularly behaviour of the intrinsic foot muscles, is however unknown.

The aim of this study was therefore to quantify EnHL for EMG signals from intrinsic foot muscles and investigate whether EMG EnHL changes during different postural tasks. If shorter CoP EnHL results from increased interference from the postural control system, such interference could relate to increased fluctuations in the excitation of the intrinsic foot muscles, reflected by shorter EMG EnHL. Therefore, shorter EnHL values are hypothesised to occur in both CoP and EMG data during postures that provide greater balance challenge (e.g. single leg or tiptoe standing). However, increased balance challenge could reduce the number of solutions available, or used by, the intrinsic foot muscle system to produce the required force vectors. This would lead to fewer fluctuations in intrinsic foot muscle excitation and hence longer EMG EnHL. Therefore, CoP motion and muscle excitation fluctuations may be decoupled, with shorter CoP EnHL associated with longer EMG EnHL.

2. Methods

2.1. Participants

Seventeen healthy participants (fourteen males and three females, age: 39.1 ± 13. 4 years, mass: 74.3 ± 14.3 kg, height: 175.7 ± 9.4 cm) voluntarily took part in the study having provided informed, written consent. This sample size is greater than previous studies of a similar nature ($N = 10$, Kelly et al, 2012; Kelly et al, 2014), and given the experimental design (three trials of three conditions), was predicted to meet the required β power of 0.8 at an α value of $p \leq 0.05$ for statistical analysis [14]. All procedures were approved by the Faculty of Science and Engineering ethics committee at Manchester Metropolitan University. Exclusion criteria for participants included self-reported foot pain or lower limb pain during the last six months.

2.2. Data acquisition

Data acquisition followed the protocol described by [3], with monopolar surface EMG recorded (2048 Hz) from the intrinsic foot muscles with a high-density 13 × 5 grid of 64 channels (2 mm diameter, 8 mm inter-electrode distance in both directions. ELSCH model, OT Bioelettronica, Turin, Italy). The skin over the plantar region of the right foot was lightly abraded with abrasive paste and cleaned to remove debris. To determine the location of the electrode grid, the adipose pads at the heel and metatarsals were palpated and the grid positioned between these regions with the columns along the longitudinal axis of the foot. The reference electrode was positioned around the right ankle.

CoP trajectories were recorded (1000 Hz) with a force plate (460 × 510 mm, Advanced Mechanical Technology, Inc., AMTI, Watertown, Massachusetts, USA) covered with a 50 mm thick Styrofoam layer to reduce electrical noise from the ground.

Each participant was instructed to perform three standing tasks: i) bipedal standing (feet flat on the ground); ii) single leg standing; iii) bipedal tiptoe (heels raised from the ground). These conditions were selected as they provided a baseline (bipedal standing) and two more challenging tasks that induce larger amplitude natural sways in both anterior/posterior (bipedal tiptoe) and medial-lateral (single leg standing) directions [9,13]. Each trial lasted 30 seconds, with participants instructed to maintain balance for the duration. Synchronisation between force plate and EMG signals was achieved using an external trigger. Participants completed three trials of each condition, presented in a random order, and all trials were taken forward for further analysis.

2.3. Data analysis

2.3.1. EMG and CoP Signal Processing

Recorded surface EMGs from the electrode array were visually inspected and channels showing noise due to poor skin-electrode interface contact or line interference were reconstructed based on the interpolation of the signals from neighbouring channels [15]. Next, wavelet analysis was used to process EMGs following the protocol provided by [16], where a filter bank of 11 wavelets was selected to represent a band-pass filter for the signal, with parameters set to ensure that the original signal intensities could be approximately reconstructed from the sum of the wavelet-transformed signals. To remove low frequency artefacts, the signal from the first wavelet was discarded so the total intensity at any given time was calculated as the sum of the intensities of 10 wavelets (frequency bandwidth: ~11 – 432 Hz). The sum of total intensity approximates the description of power [16], so half of the power can be seen as practically equal to the square of the root mean square values [16,17]. The total intensity from the wavelet transformed signal was calculated for each channel (resulting in a map of 64 EMG intensities for each signal time point) and was filtered with a high-pass bidirectional filter (2nd order Butterworth filter, cut-off 10 Hz) to remove the slower temporal signal components, leaving components relating to short-term fluctuations in the envelope profile [8]. Finally, each signal was resampled to 1000 Hz, so that the time difference between data points was 1 ms and it matched the sampling frequency of the force plate data.

The force plate-based CoP from each trial was calculated with Visual 3D (C-motion. Inc, Germantown, MD). Both medial-lateral (CoP_{ML}) and anterior-posterior (CoP_{AP}) trajectories were filtered in the same way as the EMG signals (2nd order Butterworth filter, cut-off 10 Hz).

2.3.2. Entropy Halflife Calculation

The middle twenty seconds of each time series was selected for analysis, to avoid inclusion of data points relating to initiation or termination of the task and to ensure equal numbers of data points were analysed in all trials. The following procedures were completed on each total intensity (EMG) and CoP signal (Fig. 1): A reshape scale approach was used to resample the time series to provide increasingly larger time intervals (τ) between consecutive data points, whilst preserving the number of data points [7]. The signal was standardized, to have a mean of zero and standard deviation of one, and rescaled across increasing τ timescales of 1 ms to 10 s. SampEn was calculated ($m = 1$; $r = 0.2$) for each resampled signal, using a freely available software [18]. SampEn calculates the conditional probability that a time series of m data points remains affiliated, within a tolerance r, if another data point is added ($m + 1$) [6]. Resulting values were normalised to the maximum SampEn calculated for the signal (when $m = 0$; $r = 0.2$) and EnHL identified as the timescale at which SampEn reached half the

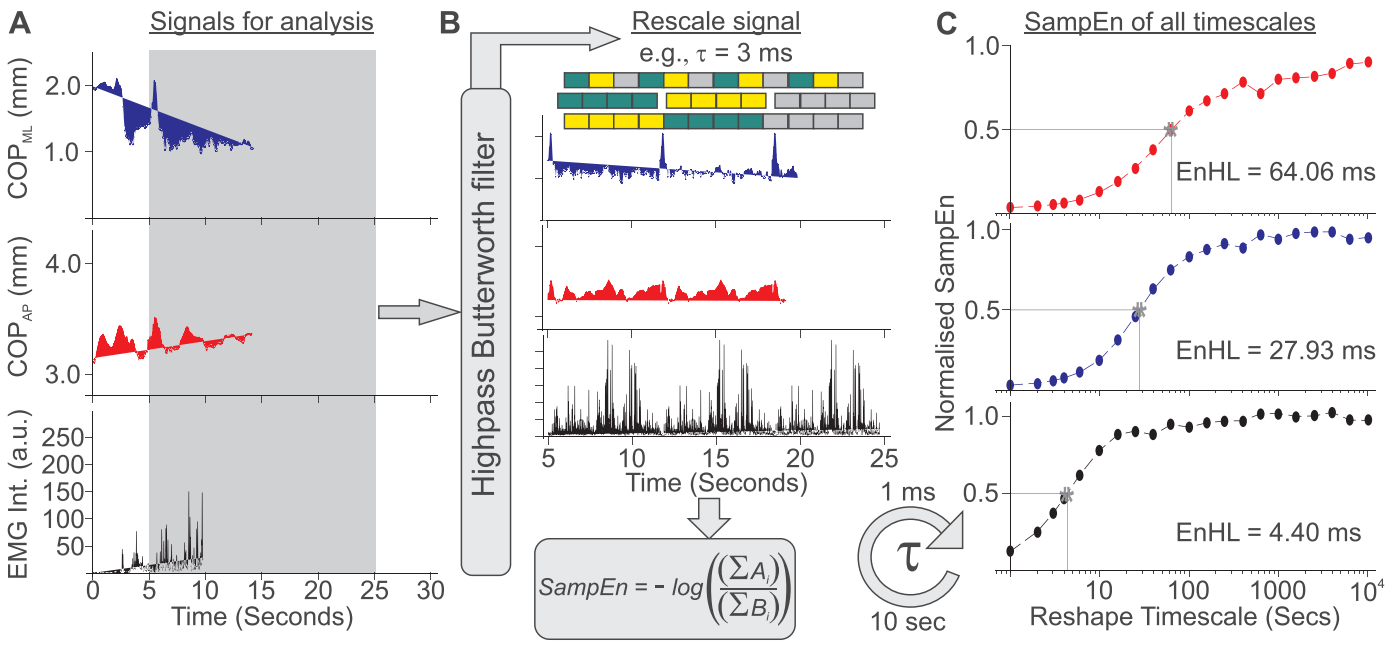

Fig. 1. Schematic overview of EnHL calculation. A) Example centre of pressure anterior-posterior (CoP_{AP}, top panel, blue) and medial-lateral (CoP_{ML}, middle panel, red) displacement and one wavelet transformed EMG (bottom panel, black), recorded during two foot tip-toe. B) Each signal undergoes a reshape scale process, illustrated for τ = 3 ms using schematic (shaded boxes) and example signals shown in A. In the schematic representation each box represents a data point with a 1 ms interval between samples. Every third value is selected to provide sub-portions of the signal that are concatenated in a randomised order[7]. The sample entropy (SampEn) of the reshaped signal is calculated, where A_i represents the count of matches with the i^{th} template of length m + 1 and B_i represents the count of matches with the i^{th} template of length m (here m = 1 and tolerance for matches r= 0.2)[6]. Each signal is reshaped over timescales (τ) ranging from 1 ms – 10 s and the resulting SampEn values (normalised to the maximum SampEn for the signal when m = 0; r = 0.2) plotted as a function of timescale (C). The timescale at which normalised SampEn = 0.5 indicates the timescale at which the signal transitions to being random (gray solid lines, each panel) and is termed the Entropy Halflife (EnHL)[7].

normalised value [7]. This value represents the timescale at which the signal transitions from containing regular, structured fluctuations to being completely random.

Preliminary inspection of the EMG EnHL values revealed that in ≈14 % of the trials (22 out of 153 trials) extremely long EnHL values occurred. These were more than 12 channels (20% of the grid), with values up to two times the order of magnitude of the other simultaneously recorded signals from the electrode grid (and values previously reported in simulated and experimental EMG data [10,11]). These values were therefore considered erroneous and related trials removed from analysis, meaning 131 trials were taken forward. From the 64 EMG EnHL values, the mean EnHL was calculated for each trial and retained for statistical analysis.

2.4. Statistical analysis

Both CoP EnHL and EMG EnHL values were tested for normality with a Kolmogorov-Smirnov normality test. These results showed that none of the EnHL results were normally distributed, but all became normal after Box-Cox transformation [19]. Linear mixed models were fit to each set of transformed EnHL values. Postural task was defined as the fixed factor, participant as the random factor with separate intercepts modelled for each participant. The location of any significant difference was identified using Bonferroni *post-hoc* analysis. For all analyses the critical factor was taken to be $\alpha \leq 0.05$. Non-transformed median ± interquartile range EnHL values are reported in the following text.

3. Results

Fig. 2 shows EMG EnHLs where values of: 18.80 ± 8.00 ms for the bipedal tiptoe; 15.46 ± 11.16 ms for single leg stance; and 9.27 ± 3.34 ms for bipedal stance were recorded. There was a statistically significant difference in EMG EnHL between bipedal stance and both single leg stance ($p < 0.001$) and bipedal tiptoe ($p < 0.001$). Fig. 3 shows the CoP EnHL values, with the lowest median values occurring during single leg stance (CoP_{ML}) and the highest during bipedal stance (both CoP_{ML} and CoP_{AP}). CoP_{ML} EnHL values were: 218.32 ± 107.24 ms for the bipedal tiptoe; 113.48 ± 83.01 ms for single leg stance; and 215.73 ± 187.58 ms for bipedal stance. CoP_{AP} EnHL values were: 146.25 ± 73.35 ms for the bipedal tiptoe; 186.43 ± 127.06 ms for single leg stance; and 259.84 ± 230.22 ms for bipedal standing. There was a statistically significant difference between each postural task for CoP_{AP} (single leg stance vs. bipedal tiptoe: p = 0.008, all other comparisons $p < 0.001$) and between single leg stance and each of the bipedal tasks for CoP_{ML} ($p < 0.001$).

4. Discussion

This study has used novel application of surface EMG electrode arrays together with information theory-based metrics to show that the patterns of intrinsic foot muscle excitations and CoP movement are significantly affected by standing posture. Significantly, changes in patterns of EMGs and CoP movement were in opposite directions. The shorter CoP EnHL for the single leg standing and tiptoe tasks indicates there is increased interference from the postural control system for these more challenging balance postures. However, the longer EMG EnHL indicate that, for the intrinsic foot muscles, there were fewer solutions available, or used, to produce the required force vectors. Taken together these results suggest that the increased fluctuations in CoP movement during balance must be predominantly driven by excitation patterns of muscles extrinsic to the foot.

The intrinsic foot muscles may have three potential roles: i) generating forces to impose motion of foot segments; ii) generating forces to stabilise the foot (e.g. stiffen the longitudinal arch and/or forefoot region); iii) intensifying sensory feedback to maintain balance (e.g.

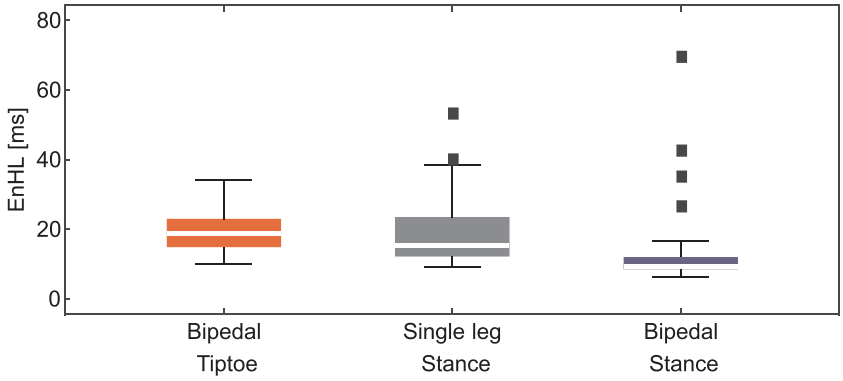

Fig. 2. Entropy Halflife values recorded for bipedal stance, single leg stance and bipedal tiptoe for EMG signal (EMG EnHL). A statistical difference was found between bipedal stance and both single leg and bipedal tiptoe conditions ($p < 0.001$).

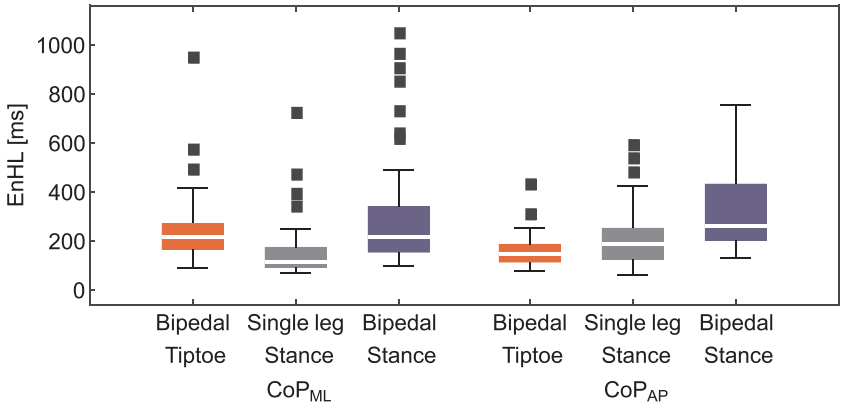

Fig. 3. EnHL distribution for CoP mediolateral trajectory (CoP_{ML}) and anterior-posterior trajectory (CoP_{AP}) for the postural tasks (bipedal tiptoe, bipedal stance, single leg stance). Significant differences in En_{HL} of CoP_{AP} occurred between each of the three conditions. Significant differences in En_{HL} of CoP_{ML} occurred between single leg standing and both bipedal tiptoe and bipedal standing conditions.

from Golgi tendon organs or muscle spindles). Previous results show that EMG EnHL increases for more challenging mechanical demands, i.e. pedalling at higher loads [8,12] or cadences [10,12], and running at faster velocities [11]. Comparing EMG EnHL for the three postural tasks studied here revealed longer EnHLs for single leg and tiptoe standing postures compared to bipedal, flat foot standing. Increased intrinsic foot muscle EMG amplitudes have previously been shown, using intramuscular recording techniques, for single leg compared to bipedal stance [20], and surface recording techniques for bipedal flat foot versus bipedal tiptoe stance [3], suggesting they respond to increased loading and act to stabilise the foot as suggested by the foot core system paradigm [21]. The longer EMG EnHLs for the more challenging balance tasks indicate that fewer solutions were available to, or used by, these muscles to solve the task demand [5,8], reflecting fewer excitation patterns used in the production of force vectors with the required magnitude and/or direction.

Fewer fluctuations in intrinsic foot muscle excitation may also facilitate an ability to provide sensory feedback. One source of muscle proprioceptive feedback is the length dependent information derived from muscle spindles. Changes in muscle activity affect muscle length and hence modulation of activity can be thought to interfere with proprioceptive feedback [22]. Fewer fluctuations in intrinsic muscle excitation may therefore enable valuable feedback from sensory organs (including Golgi tendon organs) to inform the balance control process. Further work is however required to determine the importance of such feedback mechanisms, particularly as short-term intervals in CoP movements have been shown to exhibit characteristics of open-loop control (i.e. not using sensory feedback) [23].

CoP EnHLs were longer than EMG EnHLs. Both signals were processed for the EnHL analysis in the same way (i.e. same filtering parameters and SampEn parameters). The differences therefore reflect more frequent fluctuations in the EMG not present in the longer duration fluctuations in CoP values. In addition to the differences in absolute EnHL values, differences in the direction of change of EnHLs between conditions was striking. In contrast to the EMG EnHLs, the longest CoP EnHL occurred during bipedal standing while the shortest occurred during single leg stance for CoP_{ML} and bipedal tiptoe for CoP_{AP}. This suggests that more adjustments in CoP are made along the axis about which the balance challenge is predominant, resulting in a less regular CoP displacement and shorter EnHL. These findings replicate results presented in [9,13]. The longer CoP EnHL for the bipedal standing could reflect postural sway dynamics being governed by the passive stiffness of the muscle tendon unit [24]. The shorter CoP EnHL for single leg standing and tiptoe tasks suggest more active components interfere with the CoP movement. The differences in change in EnHL between the CoP and EMG signals however indicate that increased CoP movement fluctuations were not accompanied by more intrinsic foot muscle excitation fluctuations.

The differences in EnHL change between EMG and CoP may reflect the fact that EnHL applied to CoP includes multiple motor control events, not only the muscle excitation pattern. Federolf et al., propose that shorter CoP EnHL could reflect a combination of more frequent interventions by the neuromuscular system and greater tolerance of variability in the mechanical configuration [9]. Correlations (determined by intramuscular recordings) have been found between intrinsic foot muscle EMG amplitudes and medial-lateral CoP waveforms, suggesting a role for these muscles in balance control [20]. While the CoP EnHL changes presented here agree with those from [9], the EMG EnHLs suggest that the greater CoP fluctuations for single leg and tiptoe postures are not the result of more frequent interventions via intrinsic foot muscle excitation. The evidence presented here therefore suggests these muscles play a greater role in stabilisation, potentially through stiffening of the longitudinal arch [25] or the forefoot region (as has been shown during locomotion [26]) to provide a more stable base of

support, rather than balance control.

The work presented here was conducted on healthy, adult individuals. As the techniques used were all non-invasive they can easily be applied to children or clinical populations, with impaired movement or balance skills. Control mechanisms in these populations may be additionally challenged by growth, ageing or pathology (e.g. peripheral neuropathy, sarcopenia) and EnHL could provide markers of change that may be useful for tracking motor relearning or potentially be predictive of risks of events such as falls.

5. Conclusion

This study provides preliminary baseline EMG EnHL values for the intrinsic foot muscles during postural tasks, which may be used as reference for future investigations. There are altered intrinsic foot muscle excitation patterns affected by the task challenge, with fewer excitation patterns during more challenging tasks. As more challenging balance tasks were also associated with *more* fluctuations in CoP movement patterns, the intrinsic foot muscles can be considered more as stabilisers rather than intrinsic to balance control. Further analysis should focus on the relationship between the excitation patterns from the intrinsic and extrinsic foot muscles, as this could provide insight into coordination between these two compartments for postural tasks.

Conflict of Interest

There are no conflicts of interest.

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