Enhancement of Synchronization between Physiological Signals during Exercise: A Preliminary Investigation

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Abstract—During running, interactions were considered between three physiological oscillators – the heart, breaths, and steps. During intense exercise, the oscillations of all three systems are close to regular, producing good conditions to observe and characterise synchronization. The origin, as well as any physiological significance, of synchronization between these systems during running is not fully accepted or understood. Furthermore, the impact on synchronization of controlling both breathing and step rate has not been previously reported in detail. This study aims to measure cardiolocomotor, cardiorespiratory and respiratory-locomotor synchronization during different running protocols. Breathing was controlled by taking a fixed number of steps per breath (ratios of *5:1* and *3:1*). Step rate was then guided at rates close to active heart rate, to instigate *1:1* phase-locking. Instantaneous phase difference quantified synchronization episodes. We have successfully observed all three forms of synchronization during all running protocols. Furthermore, coupling between heartbeats and steps was more pronounced when step rate was guided, and both cardiorespiratory and respiratory-locomotor coupling were extended when breathing rate was fixed to steps. These are exciting initial results from a novel experimental design, highlighting the complex interconnection that exists between these three systems during running, and the conditions to best observe the phenomena.

I. INTRODUCTION

Coupling between physiological oscillators has been observed and reported for many decades [1, 2]. Due to the nonlinear behaviour of many physiological systems, the quantification of the interaction between them can be complex, particularly when considering synchronization [3]. In weakly coupled systems, synchronization causes the adjustment of rates between the oscillators until entrainment occurs. The strongest synchronization is typically observed for *1:1* ratio between periods of the coupled oscillators. More generally, *m* periods will be observed for *n* periods of the other coupled system, or *n:m* locking during the synchronization. However, due to the stochastic nature of physiological processes, along with the presence of noise, characterising the synchronization by using a ratio between periods is difficult. Phase-locking is used instead [3] and systems are identified as synchronized assuming the relative phase difference between them does not increase by more than 2π (or $2m\pi$ in the general case).

Cardiorespiratory synchronization (CRS) and its origins have been considered extensively at rest (see [4] and references therein). During spontaneous breathing, several heartbeats occur per breath, usually around *3:1* or *5:1*, however this effect is elusive [5]. The strongest interaction between oscillators occurs when rates are equivalent [3, 6], and *1:1* phase-locking synchronization has been described in detail when breathing is elevated above heart rate [4]. During these experiments, athletes were found to have extended durations of synchronization compared to non-athletes, suggesting a predisposition to the phenomenon [7]. As the origins of CRS are suggested to be physiological or neurological, it follows that this coupling could exist during rhythmic exercise. Conversely, it has previously been reported that, although observed, CRS is less pronounced during exercise [8].

Cardiolocomotor synchronization (CLS) has previously been reported during walking, running and cycling [1, 9]. The first report dates back to 1921, and recounts a man walking up a hill, stepping in time with his pulse to avoid breathlessness [1]. Heartbeats and steps become synchronized when rates are close to each other, usually in a *1:1* ratio [9]. To force entrainment, a participant can be guided to run at a rate close to their heart rate. To this end, previous studies have used either a fixed cadence for all participants [10] or adaptive-paced metronome determined by instantaneous heart rate [11] to guide step rate and instigate *1:1* phase-locking. However, the former does not provide the bespoke rate required for an individual to perform "naturally", and the latter does not fix the rate of one oscillator – it would be difficult to quantify entrainment, as both signals fluctuate instantaneously. These methods [10, 11] produce artificial conditions rather than measuring natural running style. Again, CLS may be mechanical or neurological in origin. A direct link between the movement generator in the spinal cord and the cardiac control centre in the brain is postulated [12], however, the origins of CLS are not fully understood [13]. Due to previous studies identifying CLS from unnatural running conditions, as well as the remaining uncertainty around the physiological validity of such an interaction, it is clear that an alternative methodology is required to better illustrate synchronization.

Coupling between breathing and locomotion (**respiratory-locomotor synchronization** (RLS)) has been considered from a mechanical, physiological and neurological perspective. Mechanical constraints within the ribcage during running means breaths and steps become synchronized, as in many mammals [14]. When quadrupeds move at a fast rate, a *1:1* relationship between steps and breaths is observed, due to the mechanical force of front-limb impact on the animal's chest, or a proposed "visceral piston" through the thoracic, more

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apparent in hopping animals [15]. It has been reported that a number of different ratios for RLS can be observed within humans [14], chiefly between *2:1* and *5:1*. However previous studies are inconsistent in their quantification of synchronization [16]. Unlike quadrupeds, human breathing is not mechanically forced in time with steps, so there is the chance of a complementary explanation that coupling between breathing and locomotion could also be neurological, previously reported for rabbits [17], or due to physiological dependencies and increases in efficiency of gas exchange [18].

The aim of this research is to investigate synchronization between cardiac, respiratory and locomotor signals. Coupling of each corresponding pair will be quantified across three distinct experimental recordings. Initially, spontaneous episodes between the three systems will be sought, to determine the validity and prevalence of its existence. Then procedures from previous research concerning CRS [4] will be applied to these signals, guiding physiological rates that could instigate synchronization.

To better observe CLS and *1:1* interaction between heart rate and steps, step rate will be controlled at rates close to elevated heart rate. Breathing will then be similarly controlled, to ensure regularity of system rate close to a natural rhythm (lower ratios than *1:1* will be used). For synchronization analysis, techniques successfully used to postulate non-coincidental CRS will be applied [4]. Previous studies have been inconsistent in describing the origins of CLS [13] and with the identification of RLS [16] during running. By applying the same procedure to all forms of synchronization, more reliable results are expected. The authors are not aware of previous research analysing all three forms of synchronization simultaneously, whilst using phase-descriptive tools for quantification of duration. Furthermore, the experimental design is novel. Predicated on the assumption genuine physiological coupling of sufficient strength exists, this experimental design and analysis techniques should allow for identification of true synchronization.

II. METHODOLOGY

A. Protocol

To measure heart rate, breathing rate and step rate, a device was selected which recorded all three signals wirelessly. This allowed experiments to be completed outside, avoiding treadmills which can affect a person's natural running form and speed. The Zephyr BioHarness 3 utilises a small module fitted to a chest strap. ECG was recorded at 250Hz sampling rate, 3-axis accelerometer at 100Hz, and breathing at 25Hz. Measurements were carried out either on an athletic track or flat sports field, to ensure external factors did not affect physiological rates. Three distinct sessions were designed with specific outcomes intended, described in Table 1. In between each 5-minute interval was a 5-minute rest, allowing time for heart rate and breathing rate to return to an elevated but not exertive level. For session 2, the breathing rate was maintained by the individual based upon their step rate, by personally counting. Emphasis was placed upon breathing out coinciding with a footfall as a datum, to ensure consistency. To guide the rates in session 3, an audio metronome was used on a smartphone. The pace of the metronome was fixed using average values from session 1.

TABLE I. EXPERIMENTAL OVERVIEW

Session	Interval	Time	Rate
1	1 2 3 4	5 min	Spontaneous (aim to maintain a constant speed)
Purpose	Calculate the average step rate (**ASR**) and average heart rate (**AHR**) for each volunteer during free-paced running. Also investigate if CLS, CRS or RLS occur naturally.		
2	1 2 3 4	5 min	5 steps per breath 5 steps per breath 3 steps per breath 3 steps per breath
Purpose	To discover the impact of breathing on all forms of synchronization, breathing rate should be fixed. Rather than specifying an abstract frequency of breathing, breaths are counted in steps, to investigate how RLS affects CLS.		
3	1 2 3 4	5 min	**ASR** (**ASR** + **AHR**) / 2 **AHR** **ASR**
Purpose	If CLS occurs through entrainment and a true interaction, it follows that rates must be initially close. By fixing the step rate at average values close to spontaneous rates, the impact on heart rate (and breathing) can be assessed.		

B. Subjects and ethics

This preliminary analysis contains recordings from eleven volunteers (aged 20 – 30, 2 females). Comparisons will not be made between subgroups at this time. All were healthy and participate in frequent exercise. Not all were regular runners. The Biomedical and Scientific Research Ethics Committee at the University of Warwick provided ethical approval for this research (REGO-2013-565). Written consent was given by all volunteers following an explanation of the experimental procedure. Blood pressure was measured via a cuff before and after each experiment, and oxygen saturation was monitored during rests throughout the experiment using a finger-clip pulse oximeter, to ensure no adverse effects from exertion. Volunteers were encouraged to warmup to avoid injury.

C. Processing

The accelerometer signal and breathing waveform were first resampled using spline interpolation, to match the ECG sampling rate, following low-pass filtering to avoid aliasing. A high-pass filter removed low frequency noise and trends from all signals, a common feature of interference from the Zephyr during running. Instantaneous rates would be identified using an event-based approach [5], specifically from interpolation of the local-midpoint crossing for steps and breaths, and R-peaks for heart rate.

Data from the X-axis of the accelerometer was smoothed using a narrowband filter around the dominant signal frequency. This filtering produced a quasi-harmonic signal representative of step periods. Individual steps were identified using a local-midpoint crossing.

Similarly, a passband filter was used to smooth the breathing signal to a quasi-harmonic form. However, due to interference from movement artefacts – identified as originating from movement due to corresponding peaks in the

spectra for the three accelerometer channels – a stopband filter was first used to remove the influence of the fundamental frequency of steps (roughly half the frequency of individual step rate – a *step cycle*). These frequencies were often close to breathing frequencies. Again, a local-midpoint crossing was used to mark individual breaths.

The ECG signal required additional steps to ensure R-peaks were identified correctly. A passband filter was used to select the frequencies of the QRS complex. The resulting waveform was differentiated, to enhance areas with high rate of change, before squaring this signal. This stage removed negative values, and enhanced the maxima, each corresponding to an R-peak. A narrowband filter selecting the fundamental frequency of this signal produced a harmonic waveform, with each period representative of the duration for a single heartbeat. By calculating the timestamps of maxima for this harmonic waveform, local maxima in the de-trended ECG could be identified, corresponding to R-peaks.

C. Analysis

Instantaneous rates were calculated from the first differences of the periods identified for the three signals and are used for visualization. These events enabled instantaneous phase to be determined for the three signals. Heart rate increases by 2π between adjacent R-peaks, so the instantaneous value of phase is defined by linear interpolation. To mitigate errors from filtering in the quasi-harmonic breaths and steps signals, the Hilbert transform was used to determine the instantaneous phase for these two signals.

For a full explanation of quantifying synchronization, please see the methodology in our recent publication and references therein [4]. For this preliminary analysis, two tools have been employed – the synchrogram and instantaneous phase difference [5]. The synchrogram allows for quick identification of possible synchronization using a plot of cyclic relative phase difference, with episodes of synchronization appearing as horizontal plateaus. The number of parallel lines is equal to the ratio of locking (a single line representing *1:1*).

The instantaneous phase difference is a continuous plot, where phase locking manifests as a constant value of phase difference, seen as a horizontal line. A sweep was conducted across a range of realistic locking ratios (from *1:1* to *1:10*) to identify regions where the phase difference between systems did not exceed 2π for an extended duration, τ. To normalise τ for individuals, it was set as the time taken for completion of 10 breaths at average breathing rate during that interval. As heart rate and step rate were significantly faster than breathing rate, this value of τ ensured many beats and steps occurred within identified synchronization episodes. Each episode within any given ratio exceeding duration of τ was recorded, allowing for identification of the longest single synchronization episode, as well as summing all episodes for total synchronization time within a 5-minute interval.

III. RESULTS

Eleven volunteers have completed session 1, corresponding to 44 intervals of spontaneous running. Six of these volunteers have also completed session 2, thus 24 intervals of controlled breathing. Finally, three have completed session 3, and so there are 12 intervals of controlled step rate. Thus, three people have completed all sessions.
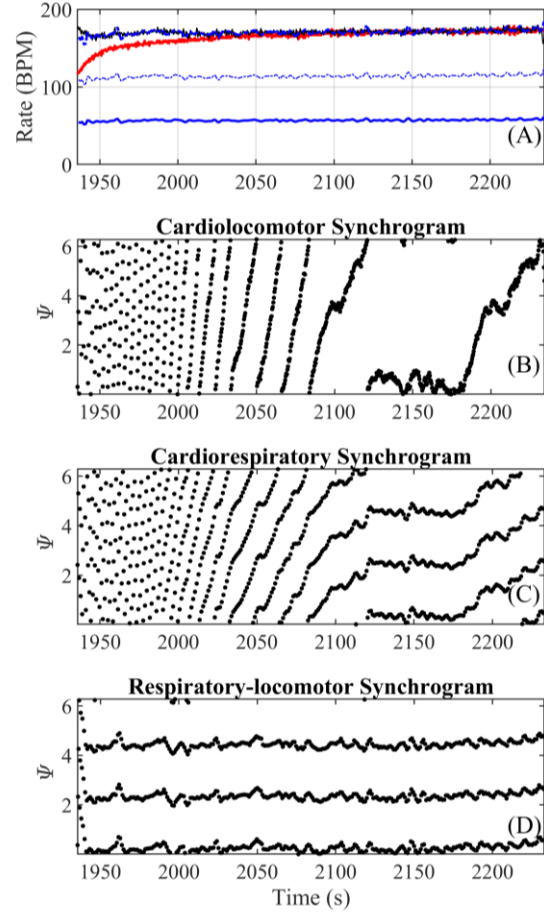

Fig 1: An interval from session 2 (*3 steps per breath*). (A) The black line shows instantaneous step rate, the red line the corresponding ramp in heart rate during this 5-minute interval. The solid blue line is the instantaneous breathing rate (around 55 BPM), while the two dashed blue lines represent 2-times (around 110 BPM) and 3-times (around 165 BPM) this breathing rate respectively. (B), (C) and (D) are synchrogram plots, titled appropriately for each of the three pairs of interactions.

All 80 intervals across the 3 sessions have been analysed. Experiments are still being conducted to produce equal group sizes across sessions. Instantaneous rates were plotted alongside the three corresponding synchrograms. An example can be seen in Figure 1. These plots are taken from an interval in session 2, where breaths are being controlled in time with a fixed number of steps. Thus, RLS at ratio *3:1* is particularly strong throughout (plot (D)). Once heart rate reaches steady state, CLS is also clearly observed, and is strongest between 2125 and 2175 seconds in plot (B). During this period, as steps and breaths are already synchronized, it is not surprising that CRS is then observed as well (plot (C)). This result is particularly interesting, as it is not clear which systems are entraining which. The additional modulation of synchronized breaths and steps during session 2 as heart rate approaches step rate appears to lead to entrainment of the heart. Instantaneous phase difference was used to determine the duration of all three types of synchronization for each interval. The longest episode was output, along with the summation of durations for all episodes within the interval, representative of total synchronized time. These durations are grouped by session, collated as boxplots in Figure 2. Durations in sessions were compared using the Wilcoxon rank-sum test (p<0.05).
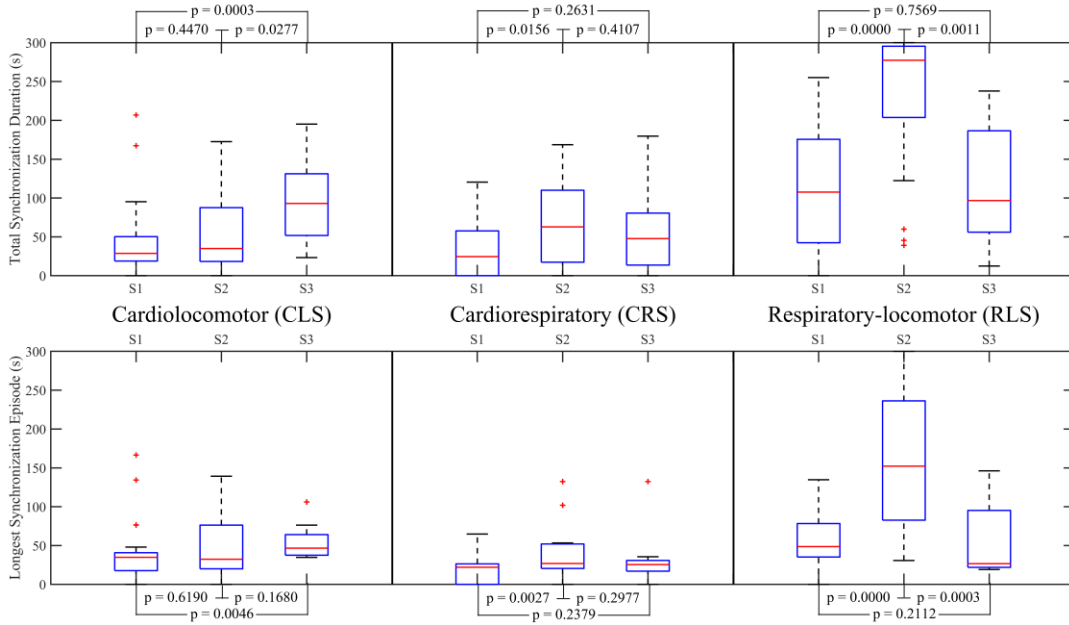
Fig 2: Durations are grouped into sessions and divided into CLS, CRS and RLS. P-values are given next to their respective comparisons (left most compares sessions 1 and 2, right most sessions 2 and 3, and top/bottom value sessions 1 and 3). **Top panel**: total duration of synchronization within an interval. All synchronization episodes are added together, assuming they exceed the stipulated 10-breath duration. **Bottom panel**: duration of longest single episode identified within an interval for each person. Only one value is considered, regardless of locking ratio, rather than longest duration per ratio observed.

All three types of synchronization can occur during any session. Not every type was observed for all volunteers in each session. All types of synchronization were less pronounced during session 1. RLS was the most pronounced during free-running conditions, likely a feature of volunteers who run regularly. CLS was identified to occur for a statistically longer total duration and longer single episode during session 3, when step rate was guided. Furthermore, all volunteers experienced at least one episode of CLS during session 3. This is in line with the intention from the experimental design. However, there is no statistical difference between CRS and RLS durations in session 1 versus session 3. Similarly, RLS total duration and single duration are statistically greater in session 2, when breathing rate is counted in time with steps, as well as there being no zero-cases. Again, this correlates with the intent from the protocol. Additionally, CRS during session 2 is also statistically more pronounced for both total duration and longest episode compared to session 1.

IV. CONCLUSIONS

This experimental protocol has successfully demonstrated the existence of all three types of synchronization between heartbeats, breaths and steps, during free-paced running. Furthermore, subsequent sessions have shown that both breathing rate and step rate can be effectively controlled to instigate synchronization for longer durations. Controlling steps at rates close to average heart rate yields significantly greater durations of CLS, while having no marked effect on breathing interactions. Controlling breathing at a fixed rate with respect to step rate increases duration of both CRS and RLS compared to session 1. As this result is not seen when comparing CRS durations for sessions 1 and 3, it would follow that controlling breathing via counting steps has a greater impact on CRS than controlling steps alone. A full comparison between sessions, as well as to previous publications, will be conducted once experiments are complete.

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