Biomedical Signal Processing and Control 54 (2019) 101601

Contents lists available at ScienceDirect

Biomedical Signal Processing and Control

journal homepage: www.elsevier.com/locate/bspc

A unified heart rate control approach for cycle ergometer and treadmill exercise

Kenneth J. Hunt*, Andreas Zahnd, Reto Grunder

Institute for Rehabilitation and Performance Technology, Division of Mechanical Engineering, Department of Engineering and Information Technology, Bern University of Applied Sciences, CH-3400 Burgdorf, Switzerland

ARTICLE INFO

Article history:
Received 22 January 2019
Received in revised form 11 May 2019
Accepted 17 June 2019

Keywords:
Heart rate dynamics
System identification
Heart rate control
Physiological control
Treadmills
Cycle ergometers

ABSTRACT

Objective: To develop a unified heart rate (HR) control approach for cycle ergometer (CE) and treadmill (TM) exercise, and to empirically compare the common controller's performance between the CE and TM.

Methods: The control method used frequency-domain shaping of the input-sensitivity function to address rejection of disturbances arising from broad-spectrum heart rate variability (HRV). A single controller was calculated using an approximate, nominal linear plant model and an input-sensitivity bandwidth specification. Fifty HR control tests were executed using the single controller: 25 healthy male participants each did one test on the CE and one on the TM.

Results: There was no significant difference in mean root-mean-square HR tracking error: 3.10 bpm $\pm$ 0.68 bpm and 2.85 bpm $\pm$ 0.75 bpm (mean $\pm$ standard deviation, bpm = beats/min); CE vs. TM; $p = 0.13$. But mean normalised average control signal power was significantly different: 1.59 bpm^2 $\pm$ 0.27 bpm^2 vs. 1.36 bpm^2 $\pm$ 0.28 bpm^2; CE vs. TM; $p = 3.5 \times 10^{-4}$.

Conclusion and significance: The lower values for RMS tracking error and control signal power for the TM point to decreasing HRV intensity with increasing HR, because, in order to match perceived exertion for the two modalities, mean HR for the TM was set 20 bpm higher than for the CE. These HR-intensity-dependent differences in HRV are consistent with previous observations in the literature. The unified HR control approach for CE and TM exercise gave accurate, stable and robust performance in all tests, thus lending support to the concept that HRV disturbance rejection is the main issue in HR control design.

1. Introduction

Feedback systems for automatic control of heart rate (HR) have been developed separately for cycle ergometers [1–3] and treadmills [4–6]. Heart rate controllers are important because they allow accurate implementation of arbitrary HR profiles, such as are employed as part of cardiovascular training programmes [7,8]; recommended strategies include high-intensity interval training (HIIT), where intensity is varied by flexibly combining exercise periods of different durations and at different levels of heart rate [9,10].

In the present work, we set out to develop and test a novel, unified heart rate control approach that can be applied to both cycle ergometers (CE) and treadmills (TM). This undertaking was motivated by the recent observation that the time constant of heart-rate dynamics at moderate-to-vigorous ("somewhat hard") exercise intensities is not significantly different for the cycle ergometer and the treadmill [11].

Notwithstanding this similarity in HR dynamics, there are three principal challenges that had to be addressed in the development of a common control design approach for the two modalities:

1. The manipulated variable is different: for the cycle ergometer, the control signal is usually a work-rate command, while for the treadmill it is a speed command. Thus, to account for the differing control-signal units, which in turn amounts to different steady-state plant gains, some form of scaling of the controller gains is required.
2. The perceived exertion of exercise at a given HR is known to be substantially higher for cycle ergometers than for treadmills [12]; and perceived exertion is similar for the two devices when HR on the CE is approximately 20 beats/min lower than on the treadmill [13]. These differences have to be accounted for in the

* Corresponding author.
E-mail address: kenneth.hunt@bfh.ch (K.J. Hunt).

https://doi.org/10.1016/j.bspc.2019.101601

Nomenclature

bpm	beats per minute
CE	cycle ergometer
CI	confidence interval
HIIT	high intensity interval training
HR	heart rate
HR*	target heart rate
HRmax	maximal heart rate
HRR	heart rate reserve
HRV	heart rate variability
k	steady-state gain
MD	mean difference
$P_{\nabla u}$	average control signal power
$P_{\nabla u'}$	normalised average control signal power
RMS	root-mean-square
RMSE	RMS tracking error
RMSSD	RMS value of differences between consecutive normal-to-normal intervals
RPE	rating of perceived exertion
rpm	revolutions per minute
SD	standard deviation
SDNN	standard deviation of all normal-to-normal intervals
τ	time constant
TM	treadmill
v	speed
v_m	mid-level speed
WR	work rate
WR_m	mid-level work rate

development of a unified CE-TM control approach, and in any comparison of control performance between the CE and TM.

3. The control approach employed must give due consideration to broad-spectrum heart-rate variability (HRV), which has been identified as the principal issue to be addressed in HR control design [6,14]. Furthermore, since the degree of HRV is dependent upon heart-rate intensity [15,16], and since, as noted above, target heart-rate levels must be set differently for the CE and TM, the impact of the HRV disturbance is likely to differ between the two modalities. These HRV-related differences must also be drawn into consideration when comparing CE and TM control performance.

The unified control design approach developed in the sequel is based upon frequency-domain shaping of the closed-loop input-sensitivity function, i.e. the transfer function between the plant HRV disturbance and the control signal [6]; the approach exploits a common HR time constant in the nominal plant model for the CE and TM [11], while using scaling to account for the difference in steady-state gains. Since the disturbance term is primarily caused by physiological HRV, appropriate shaping of the input-sensitivity function aims to ensure that the control signal is not unduly excited in frequency ranges that would be perceptible to the person performing the exercise.

The aim of this work was twofold: to develop a unified heart rate control approach for cycle ergometer and treadmill exercise; and to systematically test and compare the common controller's performance between the CE and TM using a single experimental cohort of 25 participants.

2. Methods

2.1. Control design approach

2.1.1. Plant model and controller

The common control structure for both exercise modalities comprises a nominal plant P_o, a feedback compensator C_{fb} and a reference prefilter C_{pf}, connected according to the structure shown in Fig. 1; this block diagram is typical for feedback control systems, where the prefilter is used to shape the reference response independently of the properties of the feedback loop (see e.g. [17,Ch. 8,p. 229]).

The generic controlled variable y is, in this instance, heart rate; the control signal u is either a work-rate command (CE) or speed command (TM); and there are three external inputs: reference signal r (target heart rate), disturbance d (which primarily represents heart rate variability), and measurement noise n. Throughout this work, the units of heart rate are taken to be beats/min (bpm), work rate is in W, and speed is in m/s.

The plant is described in general by the strictly-proper rational function $P_o(s) = B_o(s)/A_o(s)$ with B_o and A_o polynomials in the Laplace-transform complex variable s; for a first-order plant, P_o is parameterised by steady-state gain k and time constant τ:

$$u \to y: \quad P_o(s) = \frac{B_o(s)}{A_o(s)} = \frac{k}{\tau s + 1}. \tag{1}$$

After normalising to make the denominator monic, $B_o(s) = k/\tau$ and $A_o(s) = s + 1/\tau$. The first-order form of the general transfer function B_o/A_o was chosen based on recommendations from separate system identification studies where it was found that this simple structure gave a dynamic modelling error of less than 3 bpm [11,18]; furthermore, control design based on a first-order plant model was found to give accurate, stable and robust performance with root-mean-square tracking error less than 3 bpm [6].

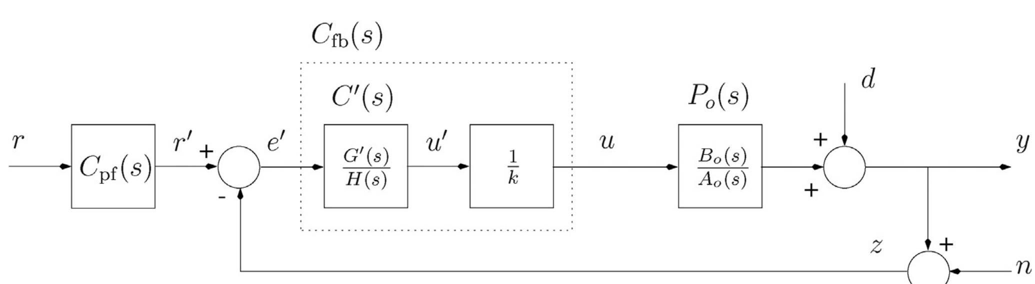


Fig. 1. Plant model and control structure. For the plant P_o, the controlled variable y is heart rate (HR), the control signal u is the manipulated variable: CE – work rate command, WR; TM – speed command, v. d is a disturbance term which, for this application, principally represents broad-spectrum heart rate variability, and n represents measurement noise. $z = y + n$ is the measurement signal. The reference signal r is the target heart rate (HR*), r' is the filtered reference, $e' = r' - z$ is the controller input, and the term u' is a notional intermediate signal. Transfer functions: P_o is the nominal plant; $C_{\mathrm{fb}} = \frac{1}{k} \cdot C'$ is a feedback compensator comprising the rational function G/H, where $G = \frac{1}{k} \cdot G'$; and C_{pf} is a reference prefilter.

The feedback compensator is the rational function $C_{\text{fb}}(s) = G(s)/H(s)$, where G and H are polynomials; C_{fb} is required here to be strictly proper (i.e. $n_g < n_h$). As noted below (Eq. (3)), G contains the factor $1/k$ and, to introduce integral action, H the factor s, thus

$$e' \to u: \quad C_{\text{fb}}(s) = \frac{G(s)}{H(s)} = \frac{\frac{1}{k} \cdot G'(s)}{sH'(s)}, \tag{2}$$

whence $G(s) = (1/k)G'(s)$ and $H(s) = sH'(s)$. The strictly-proper requirement results in both the compensator and the input-sensitivity function (U_o in Eq. (4)) having low-pass characteristics, i.e. $\lim_{\omega\to\infty} |C_{\text{fb}}(j\omega)| = 0$ and $\lim_{\omega\to\infty} |U_o(j\omega)| = 0$.

2.1.2. Control design by input-sensitivity shaping

A feedback-design approach was previously developed that gives a nominal input-sensitivity function U_o that is low-pass and of first order, whose magnitude thus decreases monotonically with frequency and is devoid of any peaking, and whose bandwidth can be specified by a single design parameter, denoted p; the compensator solution for this problem statement is [6]

$$C_{\text{fb}}(s) = \frac{\frac{1}{k} \cdot p(s + \frac{1}{\tau})}{s(s + p + \frac{1}{\tau})}, \tag{3}$$

which depends only on the specified bandwidth p and the given plant parameters k and τ. With reference to the generic form of Eq. (2), it is seen that $G'(s) = p(s + 1/\tau)$ and $H'(s) = s + p + 1/\tau$.

With the compensator Eq. (3) and the first-order plant Eq. (1), the input sensitivity function, [17,Ch. 11], for the feedback system is

$$d, r', n \to u: \quad U_o(s) = \frac{C_{\text{fb}}(s)}{1 + C_{\text{fb}}(s)P_o(s)} = \frac{\frac{p}{k}}{s + p} \tag{4}$$

which, by design, is a first-order transfer function with bandwidth p and a magnitude that monotonically decreases with frequency towards 0.

2.1.3. Common controller for CE and TM

The structure of the feedback compensator C_{fb} in Eq. (3) suggests a unified control strategy for plants which share a common time constant τ and differ only in the steady-state gain k, which is the case for the cycle ergometer and treadmill: C_{fb} contains a modality-dependent scaling factor $1/k$, but, under the assumption that the same bandwidth factor p is chosen and the two exercise modalities share a common τ, the remaining transfer function term is the same in each case. Writing the feedback compensator transfer function as $C_{\text{fb}}(s) = \frac{1}{k} \cdot C'(s)$ (see also Fig. 1), the common, "unscaled" term C' can be identified from Eq. (3) as

$$C'(s) = \frac{p(s + \frac{1}{\tau})}{s(s + p + \frac{1}{\tau})}. \tag{5}$$

The controller is then implemented according to the notional structure depicted in Fig. 1, where the gain k is set depending on the modality to either $k = k_{\text{CE}}$ (cycle ergometer) or $k = k_{\text{TM}}$ (treadmill).

Furthermore, it is noted that the notional auxiliary control signal $u' = k\,u$, that is to say, the output of C' when $1/k$ follows C' (see Fig. 1), is nominally identical for both modalities; this can be seen by considering that the controller factor $1/k$ is effectively cancelled by the term k contained in the plant numerator $B_o = k/\tau$. This observation can be used to define a common, quantitative measure for average control signal power $P_{\nabla u'}$, as described in the sequel (Eq. (15), Section 2.4).

It can be seen that the input sensitivity function U_o in Eq. (4) is also scaled by $1/k$ and is therefore different for the cycle ergometer and treadmill. A common, normalised input sensitivity function, denoted U_o', can be obtained, with reference to Fig. 1, by employing the auxiliary control signal u':

$$d, r', n \to u': \quad U_o'(s) = \frac{C'(s)}{1 + C'(s)P_o(s)} = \frac{p}{s + p}. \tag{6}$$

In a similar vein, the sensitivity and complementary sensitivity functions S_o and T_o, [17,Ch. 11], are obtained, respectively, as

$$d \to y: \quad S_o(s) = \frac{1}{1 + C_{\text{fb}}(s)P_o(s)} = \frac{s(s + p + \frac{1}{\tau})}{(s + p)(s + \frac{1}{\tau})} \tag{7}$$

and

$$r', n \to y: \quad T_o(s) = \frac{C_{\text{fb}}(s)P_o(s)}{1 + C_{\text{fb}}(s)P_o(s)} = \frac{\frac{p}{\tau}}{(s + p)(s + \frac{1}{\tau})}. \tag{8}$$

It is observed here that, by virtue of cancellation in the forward path $C_{\text{fb}}P_o$ of the terms $1/k$ in C_{fb} and k in P_o, S_o and T_o are independent of k, and are therefore the same for the cycle ergometer and the treadmill, thus obviating the need for any form of rescaling.

2.1.4. Controller calculation

In the preceding system identification study, [11], it was found that the time constant of heart-rate dynamics around "somewhat hard" exercise intensity is not significantly different for the cycle ergometer and the treadmill. The mean time constant, taken as the average across 50 individual models (25 participants and two modalities, viz. CE and TM), was found to be $\tau = 65.6$ s (range 34.3–120.2). The mean steady-state gains for the 25 CE models and 25 TM models, which have different units for each modality, were $k_{\text{CE}} = 0.392$ bpm/W (range 0.180–0.796) and $k_{\text{TM}} = 26.2$ bpm/(m/s) (range 13.3 to 62.9), thus giving the two nominal plant models

$$P_{o\text{CE}}(s) = \frac{0.392}{65.6s + 1}, \quad P_{o\text{TM}}(s) = \frac{26.2}{65.6s + 1}. \tag{9}$$

In the present work, the desired closed-loop input-sensitivity bandwidth was chosen to be 0.01 Hz, whence $p = 0.0628$ rad/s. The common compensator C' for both the cycle ergometer and the treadmill is then readily obtained from Eq. (5) using $p = 0.0628$ and $\tau = 65.6$ as

$$C'(s) = \frac{p(s + \frac{1}{\tau})}{s(s + p + \frac{1}{\tau})} = \frac{0.0628s + 0.000957}{s(s + 0.0781)}, \tag{10}$$

which was implemented as in Fig. 1, while setting $k = k_{\text{CE}}$ or $k = k_{\text{TM}}$ depending on the modality.

The corresponding input-sensitivity function is, from Eq. (6),

$$U_o'(s) = \frac{p}{s + p} = \frac{0.0628}{s + 0.0628}. \tag{11}$$

As noted above, the three sensitivity functions U_o', S_o and T_o are the same for both the cycle ergometer and the treadmill: the nominal sensitivity function magnitudes are shown in a Bode plot, where the bandwidth of each function has been highlighted (Fig. 2).

A reference prefilter C_{pf} was employed to shape the reference-tracking response independently of the disturbance rejection and measurement noise properties of the feedback loop (Fig. 1). Similar to the approach set out in Hunt and Fankhauser [6], the design goal for reference tracking was to achieve an overall reference response $r \to y$ with a specified 10 % to 90 % rise time t_r, according to a second-order transfer function T_{cl} with critical damping (i.e. damping factor $\zeta = 1$). Since the transfer function from the filtered reference signal r' to y is T_o (see Eq. (8)), C_{pf} is obtained as

$$C_{\text{pf}} = T_o^{-1} T_{\text{cl}}. \tag{12}$$

The desired rise time was set here to $t_r = 120$ s, thus giving a somewhat faster reference response than with T_o alone, which had a rise time of approximately 151 s.

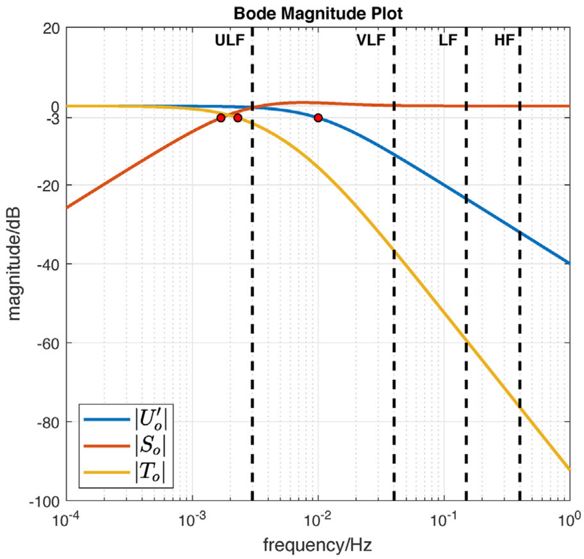


Fig. 2. Closed-loop frequency responses U_o' (normalised input sensitivity function, Eq. (6)), S_o (sensitivity, Eq. (7)) and T_o (complementary sensitivity, Eq. (8)). The red dots mark the respective −3 dB bandwidths. The vertical dashed lines bound the classical heart rate variability frequency bands: ultra-low frequency (ULF, 0.0033 Hz), very-low frequency (VLF, 0.04 Hz), low frequency (LF, 0.15 Hz) and high-frequency (HF, 0.4 Hz). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

2.2. Experimental design and test procedures

Twenty-five healthy males participated in the study. The participants were regular exercisers, non-smokers, aged between 22 years and 32 years, and had body mass from 62 kg to 114 kg, height 1.65 m to 1.93 m, and body mass index from 19.9 kg/m^2 to 34.0 kg/m^2. This same participant cohort had taken part in the preceding system identification study [11].

Each participant performed one feedback control test on the CE and one on the TM; the order of presentation of device, i.e. TM then CE vs. CE then TM, was randomised; the study thus had a repeated-measures crossover design with counterbalancing. Each test was carried out on a separate day with at least 48 h between tests. For a given period prior to each test, participants were required to avoid strenuous activity (24 h), caffeine (12 h), and large meals (3 h).

The formal protocol that was employed for feedback control tests is graphically illustrated in Fig. 3. The protocol comprised four phases: warm up, rest, formal measurement phase, and cool down. In the formal measurement phase, a target HR profile HR* was defined as a square wave with variations around a mid-level that was adjusted individually to correspond approximately to the boundary between exercise intensity levels perceived to be moderate or vigorous. The manipulated variable (CE – work rate, WR; TM – speed, v) was automatically and continuously adjusted by the feedback controller described above (Section 2.1).

The exercise intensity level was assessed using the Borg rating of perceived exertion (RPE) category rating scale [19,20], whereby the target mid-level intensity was chosen to be RPE = 13 ("somewhat hard"). In terms of HR, the boundary between moderate and vigorous intensities is defined as 76.5 % of maximal HR [8], while maximal HR can be approximated in units of beats/min (bpm) in dependence on age in years as HRmax = 220 − age [21], thus giving a mid-level heart rate HR$_m$ = 0.765 × HRmax = 0.765 × (220 − age).

When seeking to compare exercise outcomes on different devices, i.e. cycles and treadmills, it is necessary to account for differences in individual perceptions of exercise intensity for the two devices [12]. It was previously reported that, for a similar, moderate-to-vigorous level of exercise intensity, the heart rate for cycle ergometry is approximately 20 bpm lower than for a treadmill [13]; this observation was closely confirmed in the system identification study preceding the present work [11]. Thus, for the CE, the target mid-level HR$_m$ was set here to be 20 bpm lower than the TM value calculated as above, with the aim of achieving similar levels of perceived intensity for the two devices, that is to say, somewhat hard, or borderline moderate/vigorous.

During tests with the CE, participants kept their cycling cadence close to 70 rpm by observing a handlebar-mounted digital display.

All formal HR control tests had four stages (Fig. 3):

1. Warm up (10 min): the manipulated variable was manually adjusted to an individual level corresponding to comfortable, low-intensity exercise.
2. Rest (10 min).
3. Formal measurement phase (30 min): the target HR* was changed in the form of a square-wave signal for 30 min with variations around a mid-level corresponding to the individual boundary between moderate and vigorous intensity, calculated as described above; the square-wave amplitude was set to 10 bpm. Thus, for both the TM and the CE, the target HR was HR* = HR$_m$ ± 10 bpm, whereby HR$_m$ was set 20 bpm lower for the CE than for the TM.
4. Cool down (10 min): the target heart rate was set to a value 15 bpm below the lower level of the preceding square wave, i.e. to HR$_m$ − 25 bpm.

2.3. Equipment and data collection

Both the treadmill (model Venus, h/p/cosmos Sports & Medical GmbH, Germany) and the cycle ergometer (model LC7, Monark Exercise AB, Sweden) were PC-controlled in real time using Matlab/Simulink (The Mathworks, Inc., USA). For testing on both devices, heart rate was obtained using a chest belt sensor (model T34, Polar Electro Oy, Finland) and a receiver module (Heart Rate Monitor Interface [HRMI], Sparkfun Electronics, USA) that was connected to the Matlab/Simulink-based controller using a real-time serial interface with a sampling rate of 1 Hz. The controller was discretised to run with a sample period of T_s = 5 s, thus the HR data were downsampled by averaging five individual values over each sample interval.

2.4. Outcome measures and statistical analysis

Accuracy of heart-rate tracking was quantitatively assessed using the root-mean-square tracking error RMSE:

$$\mathrm{RMSE} = \sqrt{\frac{1}{N}\sum_{i=1}^{N}(\mathrm{HR_{nom}}(i) - \mathrm{HR}(i))^2}, \tag{13}$$

where i are discrete time indices (with T_s = 5 s) during the evaluation period and HR$_{\mathrm{nom}}$ is the nominal heart rate response obtained by simulating the nominal closed-loop transfer function T_{cl}.

The intensity of the manipulated variable was quantified using the average control signal power, defined formally as the average power of changes in the manipulated variable, denoted $P_{\nabla u}$. Here, the manipulated variable has the generic notation u, which refers specifically to speed v for the TM and to work rate WR for the CE. Thus,

$$P_{\nabla u} = \frac{1}{N-1}\sum_{i=2}^{N}(u(i) - u(i-1))^2. \tag{14}$$

This variable has different units for the two modalities: for the CE, the units of $P_{\nabla u}$ are W^2, and for the TM the units are m^2/s^2. A com-

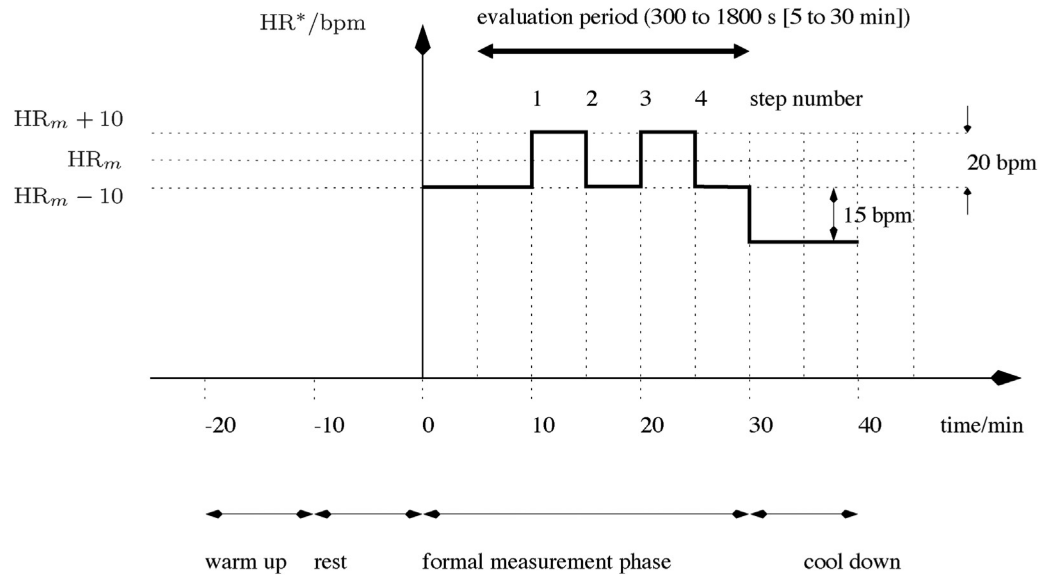


Fig. 3. Feedback control test protocol for the cycle ergometer and treadmill: target heart rate profile HR*; HR_m is the mid-level heart rate during the formal measurement phase. Note that HR_m was set 20 bpm lower for the CE than for the TM in order to achieve similar levels of perceived exertion.

mon, normalised average control signal power with the same units can be derived using the auxiliary control signal u': since $u' = ku$, and because the units of k for the CE and TM are bpm/W and bpm/(m/s), respectively, it follows that the units of u' are bpm in both cases. Thus, the average power of changes in u', i.e. the normalised average control signal power $P_{\nabla u'}$, is:

$$P_{\nabla u'} = \frac{1}{N-1}\sum_{i=2}^{N}(u'(i) - u'(i-1))^2 = k^2 P_{\nabla u}. \quad (15)$$

This normalised variable has the advantage that the units for both modalities are the same (bpm^2), thus facilitating direct comparison between the CE and TM.

The outcomes RMSE and $P_{\nabla u'}$ were calculated over an evaluation period from 300 s to 1800 s of the formal measurement phase (Fig. 3).

During the formal measurement phase, the individual perception of exercise intensity was manually recorded using the Borg RPE scale at the four time points occurring one minute before the end of each phase of step change in the target HR (Fig. 3): 14 min (840 s), 19 min (1140 s), 24 min (1440 s) and 29 min (1740 s); the definitive RPE was taken as the average of these four values.

Hypothesis testing was carried out to check for differences in outcomes between the CE and TM. Prior to testing, normality of sample differences was assessed using the Kolmogorov-Smirnov test with Lilliefors correction. Paired two-sided t-tests were employed for normal data and Wilcoxon signed rank tests otherwise. The significance level was set as $\alpha = 0.05$. Statistical calculations were done using the Matlab Statistics and Machine Learning Toolbox (The Mathworks, Inc., USA) and R (R Foundation for Statistical Computing, Austria).

3. Results

A selection of original data records from feedback control tests is provided for both the cycle ergometer and the treadmill (Fig. 4): for each modality, measurements for the tests having the lowest, median and highest values for RMS tracking error amongst all participants are shown. Within the figure, each subplot includes two panels: the upper panel displays the target, measured and simulated heart rate signals; the lower panel shows the control signal, which for the cycle ergometer is the work rate and, for the treadmill, the speed.

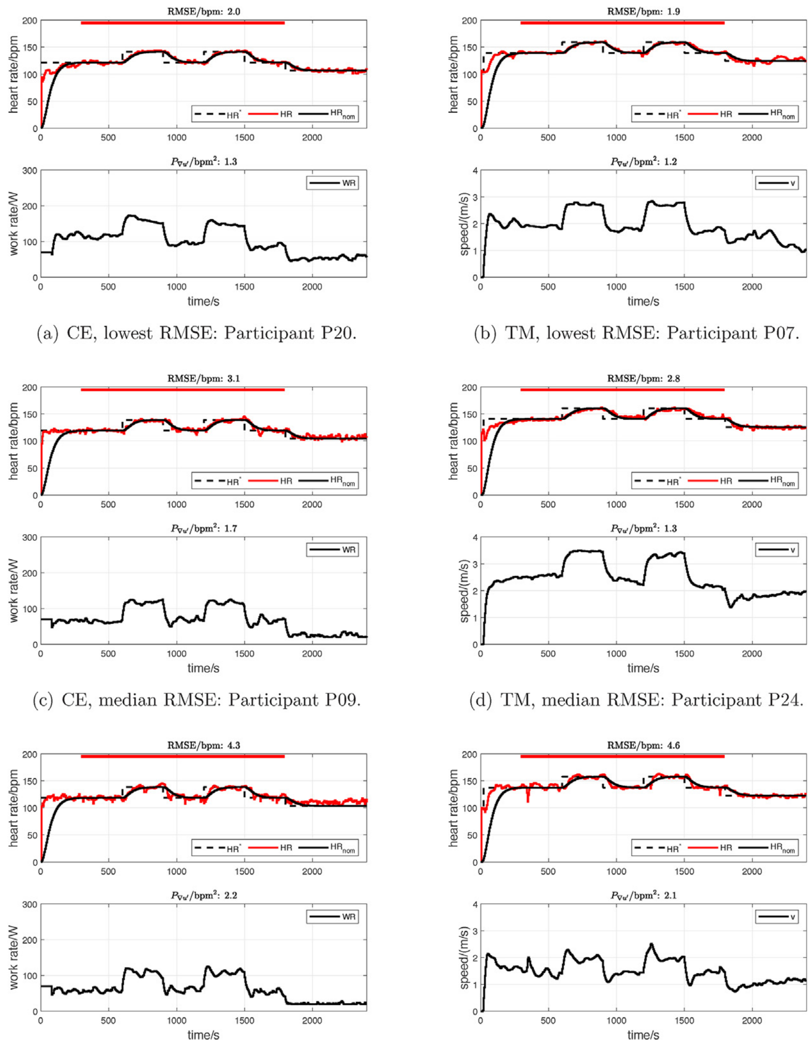
Overall, there was no significant difference in heart rate tracking accuracy between the cycle ergometer and treadmill: root-mean-square tracking error RMSE was 3.10 bpm ± 0.68 bpm and 2.85 bpm ± 0.75 bpm (mean ± standard deviation), respectively, with $p = 0.13$ (Table 1). But the normalised average control signal power $P_{\nabla u'}$ was significantly different for the two modalities: 1.59 bpm^2 ± 0.27 bpm^2 vs. 1.36 bpm^2 ± 0.28 bpm^2; CE vs. TM; $p = 3.5 \times 10^{-4}$; Table 1. A graphical illustration of the statistical comparison of means is given in Fig. 5, with RMSE in Fig. 5(a) and $P_{\nabla u'}$ in Fig. 5(b). These plots show the dispersion of the individual samples, mean values and confidence intervals (CIs). The latter allow the significance (or otherwise) of differences between the means to be visually ascertained: when a significant difference exists, the value 0 will lie outwith the respective CI.

Both the cycle ergometer and the treadmill gave very similar levels of perceived intensity close to the "somewhat hard" value of 13: RPE was 12.6 ± 1.1 vs. 12.5 ± 1.4, CE vs. TM, $p = 0.38$ (Table 1). In line with the setting of the target mean HR for the cycle ergometer to be 20 bpm lower than for the treadmill, with the aim of achieving similar levels of perceived exertion, actual mean HR for the CE was exactly 20 bpm lower: 127.0 bpm vs. 147.0 bpm (target mean HR was 126.7 bpm vs. 146.7 bpm, CE vs. TM).

Table 1
Outcomes for cycle ergometer vs. treadmill and p-values for comparison of means (see also Fig. 5).

	Mean ± SD		MD (95 % CI)	p-value
	CE	TM	CE − TM	
RMSE/bpm	3.10 ± 0.68	2.85 ± 0.75	0.26 (−0.08, 0.59)	0.13
$P_{\nabla u'}$/bpm^2	1.59 ± 0.27	1.36 ± 0.28	0.24 (0.13, 0.37)	3.5×10^{-4}
RPE/(6–20)	12.6 ± 1.1	12.5 ± 1.4	0.15 (−0.2, 0.5)	0.38

$n = 25$; CE: cycle ergometer; TM: treadmill; SD: standard deviation; MD: mean (RMSE, RPE) or median ($P_{\nabla u'}$) difference of CE - TM; 95 % CI: confidence interval for the mean (RMSE, RPE) or median ($P_{\nabla u'}$) difference; p-values: paired two-sided t-tests (RMSE, RPE) or Wilcoxon signed-rank test ($P_{\nabla u'}$); RMSE: root-mean-square error; $P_{\nabla u'}$: normalised average control signal power; RPE: rating of perceived exertion (Borg scale); bpm: beats per minute.



(a) CE, lowest RMSE: Participant P20.

(b) TM, lowest RMSE: Participant P07.

(c) CE, median RMSE: Participant P09.

(d) TM, median RMSE: Participant P24.

(e) CE, highest RMSE: Participant P13.

(f) TM, highest RMSE: Participant P19.

Fig. 4. Results with cycle ergometer (CE, left column of graphs) and treadmill (TM, right column) with the lowest (a, b), median (c, d) and highest (e, f) values for RMS tracking error amongst all participants. In the upper part of each figure, HR* is the heart rate reference, HR is the measured heart rate, and HR_{nom} is the nominal, simulated heart rate. The lower graphs show the manipulated variable: CE – work rate command, WR; TM – speed command, v. The thick red horizontal bars mark the overall outcome evaluation interval $300 \leq t \leq 1800$ s. RMSE: root-mean-square tracking error, Eq. (13). $P_{\nabla u'}$: normalised average control signal power, Eq. (15). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

4. Discussion

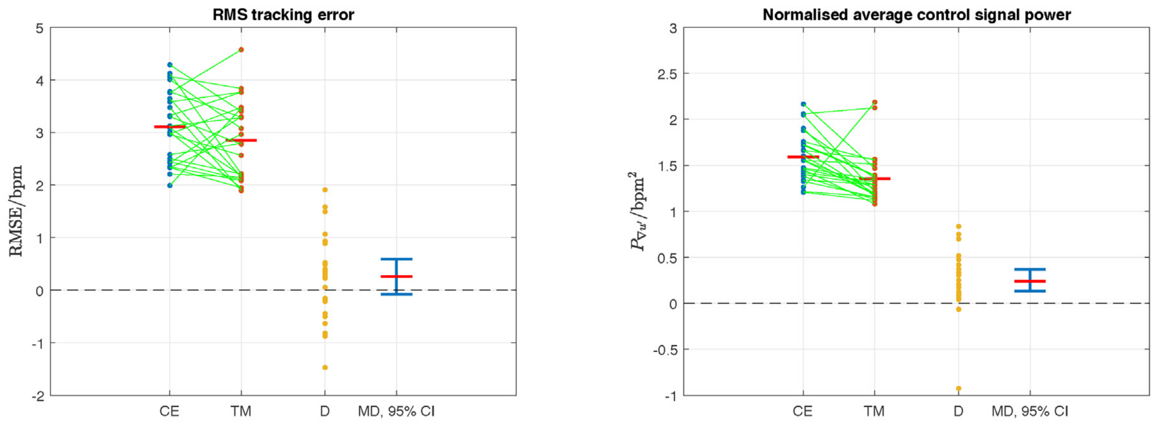
The aim of this work was to develop a unified heart rate control approach for cycle ergometer and treadmill exercise, and to empirically test and compare the common controller's performance between the CE and TM.

The unified control design strategy derived a common controller for the two modalities, where the controller transfer function is dependent upon the common plant time constant and on a single design parameter, namely the desired bandwidth of the closed-loop input-sensitivity function; the controller merely requires to be scaled according to the differing steady-state gains for the CE and TM. This design approach, and specifically the bandwidth design parameter p, allows a direct tradeoff to be set between tracking accuracy and control signal intensity, both of which are directly influenced by broad-spectrum heart rate variability.

For the experimental evaluation, a single controller was applied in 50 HR-control experiments, i.e. 25 participants each did one test on the CE and one test on the TM. For controller calculation, the plant time constant was taken from a previous study as the average from 50 individual system identification experiments (25 on the CE and 25 on the TM), while the plant steady-state gains for CE and TM were averages from the 25 respective identified values [11]. It should be noted that, although average values were taken for the nominal plant parameters, the individually-identified values varied on a wide range (see Section 2.1.4): the 50 time constants were on the range 34.3–120.2 s, the 25 CE gains ranged from 0.180 bpm/W to 0.796 bpm/W, and the 25 TM gains from 13.3 bpm/(m/s) to 62.9 bpm/(m/s).

The controller gave highly accurate tracking performance with mean root-mean-square tracking error RMSE around 3 bpm, and with no significant difference between the CE and TM. The control signal was found to be smooth and stable in all tests (see, for example, the "best," median and "worst" results in Fig. 4). Normalised average control signal power $P_{\nabla u'}$ was significantly lower for the TM: as described below, this may reflect a lower intensity of heart rate variability, secondary to a higher heart rate, on the TM.

The single feedback controller was not calculated specifically for any of the participants tested, but the controller nevertheless gave accurate and stable performance in 50 individual experiments involving 25 participants and two different exercise modalities. Given the very wide ranges for the individually-identified plant



(a) Root-mean-square tracking error, RMSE. (b) Normalised average control signal power, $P_{\nabla u'}$.

Fig. 5. Primary outcomes: data samples for RMSE and $P_{\nabla u'}$ for all 25 participants for the cycle ergometer CE and treadmill TM (see also Table 1). The green lines link the sample pairs from each participant; for the individual samples, the red horizontal bars depict mean values (given numerically in Table 1). D = CE − TM is the difference between the paired samples. MD is the mean (RMSE) or median ($P_{\nabla u'}$) difference (red horizontal bar), with its 95 % confidence interval (CI) in blue. For RMSE, the value 0 is within the 95 % CI, indicating no significant difference between the means: this conforms with $p > 0.05$ for this variable (Table 1). For $P_{\nabla u'}$, 0 is outwith the 95 % CI, indicating a significant difference ($p < 0.05$, Table 1). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

time constants and steady-state gains, this provides strong empirical evidence of robustness of the input-sensitivity design approach, and underscores the observation made previously that dealing with the broad-spectrum HRV disturbance is the principal design challenge for HR control, while issues of parametric and structural plant uncertainty play a secondary role [6].

Differences in the observed values of $P_{\nabla u'}$ and RMSE can be interpreted in the light of known, intensity-dependent changes in heart rate variability. This interpretation is possible because: (i) it has previously been observed that the intensity of HRV decreases with increasing heart rate [15,16]; and, as detailed below, (ii) the two outcomes $P_{\nabla u'}$ and RMSE are closely related to standard time-domain-based HRV measures [22,23]. As noted in the Results, mean HR for the treadmill was 20 bpm higher than for the cycle ergometer as a consequence of matching the perceived exertion for the two modalities. Thus, HRV intensity would be anticipated to be lower for the treadmill. Analysis of $P_{\nabla u'}$ and RMSE shows that this was indeed the case in this study:

1. The normalised average control signal power $P_{\nabla u'}$ was found to be significantly lower for the treadmill than for the cycle ergometer (mean values 1.36 bpm^2 and 1.59 bpm^2, respectively, $p = 3.5 \times 10^{-4}$; Table 1). Furthermore, it is noted that the functional form of $P_{\nabla u'}$, Eq. (15), is similar to the definition of the standard time-domain HRV measure RMSSD (root-mean-square value of differences between consecutive normal-to-normal intervals, [22,23]).
2. In a similar vein, RMSE was somewhat (albeit not significantly) lower for the treadmill than for the cycle ergometer (mean values 2.85 bpm and 3.10 bpm, respectively, $p = 0.13$; Table 1). The functional form of RMSE, Eq. (13), is close to the standard time-domain HRV measure SDNN (standard deviation of all normal-to-normal intervals, [22,23]).

Taken together, these results may be a reflection of the higher treadmill HR and concomitant lower HRV. This provides further evidence that HRV intensity decreases with increasing HR and is consistent with previous observations [15,16].

With regard to the perception of exertion and its relation to HR, it is noted that, for the CE, the observed mean HR of 127.0 bpm is close to the HR that nominally corresponds to an RPE of 13, that is to say, a HR of 130 bpm: the Borg RPE scale is designed to linearly increase by a factor of 10 in relation to HR for cycle ergometer exercise. The observed mismatch between RPE and HR for the TM (mean HR for the TM was 147.0 bpm) is likely a consequence of the Borg scale having been designed specifically for cycle-ergometer exercise: as stated by Borg, [24], "The scale was designed to grow linearly with exercise intensity and heart rate for work on the bicycle ergometer."

The practical significance of this work lies in the potential application to exercise testing and prescription [8], and also in the realm of cardiac rehabilitation [25]. In these applications, heart rate provides a straightforward means to characterise exercise intensity, while feedback control allows specific HR profiles to be followed with precision.

In conclusion, a unified HR control approach was developed for cycle ergometer and treadmill exercise. The approach allowed a single controller to be calculated using an approximate, nominal linear plant model and a common input-sensitivity bandwidth specification. The controller gave accurate, stable and robust performance in an experimental series with 25 participants exercising on both modalities. These results support the concept that HRV disturbance rejection is the main issue in HR control design.

Ethics statement

The research protocol for this study was reviewed and approved by the Ethics Committee of the Swiss Canton of Bern (Ref. 2017-01894). Written, informed consent was obtained from all participants.

Author contributions

KH designed the study. AZ and RG did the data acquisition. AZ, RG and KH contributed to the analysis and interpretation of the data. KH wrote the manuscript; AZ and RG revised it critically for important intellectual content. All authors read and approved the final manuscript. AZ and RG contributed equally to this work.

Declaration of interest

None.

Funding disclosure

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

Data availability statement

The raw data supporting the conclusions of this manuscript will be made available by the authors, without undue reservation, to any qualified researcher.

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