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Youri Duchene, Frédéric Simon, Geoffrey N Ertel, Hugo Maciejewski, Gérome C Gauchard, et al.. The stroke rate influences performance, technique and core stability during rowing ergometer. Sports Biomechanics, 2024, 10.1080/14763141.2024.2301992 . hal-04402123

**HAL Id: hal-04402123**

**<https://hal.univ-lorraine.fr/hal-04402123v1>**

Submitted on 18 Jan 2024

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**Title: The stroke rate influences performance, technique and core stability during rowing ergometer.**

Authors:

Youri Duchene<sup>ab\*</sup>, Frédéric R. Simon<sup>abc</sup>, Geoffrey N. Ertel<sup>ab</sup>, Hugo Maciejewski<sup>d</sup>, Gérome C. Gauchard<sup>abc</sup> and Guillaume Mornieux<sup>abc</sup>

<sup>a</sup>*Université de Lorraine, DevAH, F-54000 Vandoeuvre-lès-Nancy, France*

<sup>b</sup>*Université de Lorraine, CARE Grand Est, Nancy, France*

<sup>c</sup>*Université de Lorraine, Faculty of Sport Sciences, Villers-lès-Nancy, France*

<sup>d</sup>*Fédération Française d’Aviron, Nogent-sur-Marne, France*

Author for correspondence:

Youri Duchene

EA 3450 Développement, Adaptation et Handicap (DevAH), Faculté de Médecine

9 avenue de la Forêt de Haye CS 50184

54505 VANDOEUVRE LÈS NANCY

Tel.: +33 (0)3 83 68 37 45

Email: [youri.duchene@univ-lorraine.fr](mailto:youri.duchene@univ-lorraine.fr)

Authors ORCiDs and media:

Youri Duchene. ORCiD: <https://orcid.org/0000-0002-2774-2333>; Twitter: @DucheneYouri.

Frédéric R. Simon. ORCiD: <https://orcid.org/0000-0001-8137-0262>

Geoffrey N. Ertel. ORCiD: <https://orcid.org/0000-0003-0483-0135>

Hugo Maciejewski. ORCiD: <https://orcid.org/0000-0003-0686-921X>

Gérome C. Gauchard. ORCiD: <https://orcid.org/0000-0002-2489-9343>

Guillaume Mornieux. ORCiD: <https://orcid.org/0000-0001-5756-9642>

1 **Abstract:**

2 The aim of this study was to determine the effect of stroke rate on performance, technique and  
3 core stability during rowing ergometer. Twenty-four high-level rowers performed maximal  
4 intensity one-minute bouts at 20, 28 and 34 spm on a RowPerfect3 ergometer. Power at the  
5 handle, legs, trunk and arms levels were determined and core kinematics and neuromuscular  
6 activations were measured. The power at the handle was enhanced with a higher stroke rate in  
7 the first half of the drive phase due to higher segment's powers. This resulted in technical  
8 changes, as for instance greater mean to peak power ratio at each segment level. The higher  
9 trunk power preceded a delayed trunk extension but without significant increase in the erector  
10 spinae activation. This underlines the role of the core stability to transfer forces at a higher  
11 stroke rate. However, no co-activation parameters between trunk flexors and extensors helped  
12 further to understand this force transfer. Rowing at low stroke rate can be a training strategy  
13 to work on earlier trunk extension, while maintaining erectors spinae levels of activation.  
14 Training at higher stroke rate will induce a rowing technique closer to competition with  
15 greater neuromuscular activations, and maximise power production.

16

17 **Keywords:** Power production - Trunk kinematics - Pelvis kinematics - Neuromuscular  
18 control - Muscles co-activations

## 19 **Introduction**

20 In rowing, stroke rates can range from 18 strokes per minute (spm) during some training  
21 sessions to rates over 40spm during a 2000m race (Hofmijster et al., 2007, 2009; Kleshnev,  
22 2016). Moreover, the stroke rate also varies during the race, with for instance values between  
23 34 to 48spm for the M1x finalists at the 2022 World Rowing Championships in Racice in the  
24 Czech Republic (Kleshnev, 2022).

25 The mean power output at the handle, main performance parameter in ergometer rowing  
26 (Bourdin et al., 2004, 2017; Buckeridge et al., 2016B; Colloud et al., 2006; Greene et al.,  
27 2013; Kleshnev, 2016) is increased with the stroke rate rise (Buckeridge et al., 2012; Held et  
28 al., 2020; Hofmijster et al., 2007; Kleshnev, 1996; McGregor et al., 2004), ranging for  
29 example from 254W per cycle at 20spm to 473W per cycle at a competitive stroke rate  
30 (Buckeridge et al., 2012). Furthermore, the rowing technique also changes with the stroke rate  
31 to allow this increase in power production. Indeed, the stroke rate has an impact on time to  
32 peak or mean to peak power (Holt et al., 2020; Kleshnev, 2016; Smith & Draper, 2006;  
33 McGregor et al., 2004), lower limb and spine joints kinematics (McGregor et al., 2004;  
34 Buckeridge et al., 2016A; Li et al., 2020) or power patterns at the handle, legs, trunk or arms  
35 level (Ettema et al., 2022; Kleshnev, 1996). However, given that the stroke rate manipulation  
36 modifies the distribution between the drive and the recovery times (Buckeridge et al., 2014;  
37 McGregor et al., 2004), these results from the different studies in the literature should be  
38 interpreted with caution. Therefore, the effect of the stroke rate on the rowing technique  
39 should be analysed on the drive phase only (Ettema et al., 2022).

40 With its central position within the kinetic chain from the legs to the arms, the trunk plays a  
41 major role in rowing performance. Indeed, the trunk is responsible for about 30% of the  
42 handle power (Kleshnev, 2000). With the increase of the stroke rate (ranging from 17 to  
43 36spm), greater spinal forces and larger ranges of motion (ROM) of the pelvis, the lumbar

44 and the thoracic spine angles were noticed (Buckeridge et al., 2016A; Li et al., 2020;  
45 McGregor et al., 2004), underlining core stability's adaptations. Core stability refers here to  
46 the ability of the trunk, with respect to the pelvis, to produce power and to transfer efficiently  
47 forces from lower limbs to the arms (Kibler, Press & Sciascia, 2006). In rowing, the core  
48 might act as a power producer (Kleshnev, 2000) and has to efficiently transfer forces from the  
49 legs to the arms. With the stroke rate increase, the role of the core in transferring the force  
50 could be enhanced as the legs produce more power (Held et al., 2020). However, its function  
51 as a power producer could also be greater as its kinematics varies with higher stroke rates.  
52 Regarding the neuromuscular aspect of the core, an early activation of spinal and gluteus  
53 muscles, followed by a relatively short co-activation phase between abdominals and spinal  
54 muscles, and then a late activation of abdominis and external oblique muscles have been  
55 described along the drive phase (Pollock et al., 2009). The co-activation between the spinal  
56 muscles and the abdominals could lead to an increase in spine stiffness (Cholewicki et al.,  
57 1997) and to a better force transfer from the lower limbs to the arms. However, the influence  
58 of stroke rate on the core neuromuscular activity remains unknown. A deeper understanding  
59 of this relationship could lead to better adaptation of core muscle training. In light of these  
60 information, it is plausible that stroke rate manipulation could influence core muscle co-  
61 activation together with core stability kinematics, and thus performance.

62

63 The purpose of this study was to determine the effect of the stroke rate on rowing ergometer  
64 performance, technique and core stability. We hypothesised that i) the production of power at  
65 the handle and at each body segments would be greater through the drive phase for the higher  
66 stroke rate condition, ii) the core kinematics would vary through the drive phase inducing a  
67 larger ROM for the higher stroke rate condition, iii) the core muscle activity and the amount  
68 of co-activation would be larger for the higher stroke rate condition.

69 **Methods**

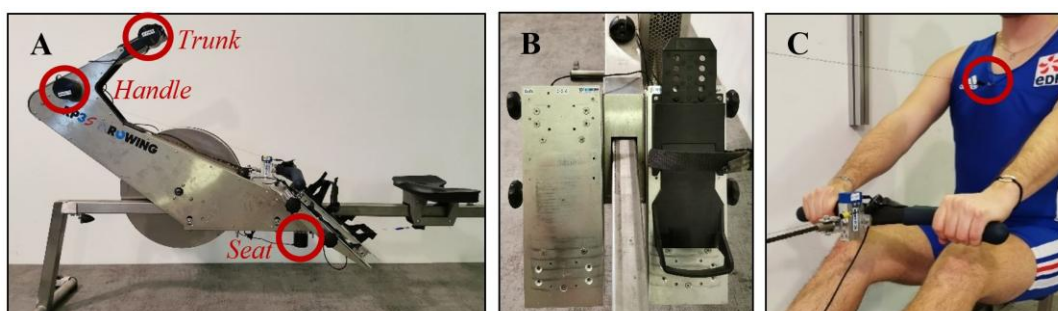
70 ***Participants***

71 Twenty-four rowers with international or national experience were recruited (17 men and 7  
72 women;  $20.8 \pm 2.8$  years old;  $1.84 \pm 0.07$ m;  $77.6 \pm 6.1$ kg;  $6\text{min } 34\text{s} \pm 30\text{s}$  on 2000m rowing  
73 ergometer performance). Participants who had current low back pain or any injury were not  
74 included in the study. The study was approved by the ethics committee Sud-Ouest et Outre  
75 Mer III (approval reference 20.07.21.43000-ID\_9115) and was conformed to requirements  
76 stipulated in the Declaration of Helsinki. All participants were informed about the procedures,  
77 purpose and possible risks associated with the experimental set-up and gave their written  
78 consent prior to testing.

79  
80 ***Test procedure***

81 After a free warm-up of 10 minutes, each rower performed three one-minute bouts at 20, 28  
82 and 34spm. A trial at their self-chosen optimal stroke rate was performed to normalise EMG  
83 signals. The instruction was to row at maximum intensity for each stroke rate.

84 The participants rowed on a mobile rowing ergometer (RP3®, Care RowPerfect BV,  
85 Haaksbergen, The Netherlands), equipped with custom-made load cells with strain gauges at  
86 left and right foot-stretchers and at the pull handle cable (BioRow Tech, BioRow, London,  
87 UK), to determine the applied effective forces. Moreover, seat, trunk, and handle  
88 displacements were measured using three position sensors (BioRow Tech, BioRow, London,  
89 UK) (Figure 1). These data were sampled at 25 Hz.



91 Figure 1: Rowing ergometer (RP3) equipped with the BioRow Tech position sensors (A,  
92 circled), and force sensors (B for the feet and C for the handle). Both foot force sensors were  
93 covered by the usual foot fixation (B, right foot).  
94

95 Core stability's sagittal kinematics was recorded with two inertial measurement units (STT-  
96 IWS, STT Systems, San Sebastián, Spain) placed in the back at the T6 vertebrae level  
97 (between the two scapulae) and between the two posterior superior iliac spines, measuring  
98 respectively the movement of the trunk and the pelvis. These two sensors did not cause any  
99 discomfort to the athlete, given their small volume (56 x 38 x 18 mm) and low mass (0.046  
100 kg). This system had a static pitch accuracy less than  $0.5^\circ$  around medio-lateral axis and a  
101 sampling frequency of 100 Hz.

102 Surface electromyography (EMG) recordings of core muscles were obtained from the rectus  
103 abdominis (RA), the external obliquus (EO), the erector spinae (ES), the tensor fasciae latae  
104 (TF) and the gluteus maximus (GM), bilaterally (Trigno™ Snap Lead wireless sensors,  
105 Delsys, Natick, MA, USA). After the skin had been shaved, sanded and cleaned, to ensure an  
106 impedance below 5 k $\Omega$ , electrodes (Ambu® BlueSensor SP, Ambu A/S, Ballerup, Denmark)  
107 were attached parallel to the muscle fibers in line with the SENIAM project's  
108 recommendations (Hermens et al., 2000). RA electrodes were placed 2 cm laterally from the  
109 umbilicus, EO electrodes at 15 cm laterally from the umbilicus, ES electrodes at about 2 cm  
110 of the L1 spinous process, TF electrodes on the line from the anterior spina iliaca superior to  
111 the lateral femoral condyle in the proximal 1/6 and GM electrodes were placed at 50% on the  
112 line between the sacral vertebrae and the greater trochanter. These EMG signals were sampled  
113 at 2000 Hz and synchronised with the motion analysis system.

114

### 115 *Data analysis*

116 Seat, trunk and handle positions as well as foot-stretcher and handle effective forces were  
117 filtered with a 4<sup>th</sup> order Butterworth low-pass filter at 5 Hz and 10 Hz, respectively (Fohanno

118 et al., 2015; Greene et al., 2009). After the computation of handle, seat and trunk speed from  
119 their measured position, the power at each level was determined as time-series data as follows  
120 (Kleshnev, 2000):

121  $P_{\text{handle}} (W) = \text{handle force} * \text{handle speed}.$

122  $P_{\text{legs}} (W) = (\text{left foot-stretcher force} + \text{right foot-stretcher force}) * \text{seat speed}.$

123  $P_{\text{trunk}} (W) = \text{handle force} * (\text{trunk speed} - \text{seat speed}).$

124  $P_{\text{arms}} (W) = \text{handle force} * (\text{handle speed} - (\text{trunk speed} - \text{seat speed})).$

125 The mean powers were then calculated over the whole drive phase. Time to peak power (T2P,  
126 %) from the catch was calculated for each segment as the occurrence of peak power delivery  
127 over the normalised drive time. Moreover, the mean to peak power ratio (M2P) was  
128 calculated for legs, trunk and arms by dividing the mean power by the peak power. The  
129 rationale for M2P was the potential relationship between the power curve shape and  
130 performance (i.e. the flatter the better) (Smith & Draper, 2006). The contribution of each  
131 segment was computed based on the mean power of the segment divided by the sum of all  
132 three segments mean power, multiplied by 100.

133 Trunk and pelvis spatial orientations were filtered with a 4<sup>th</sup> order Butterworth low-pass filter  
134 at 10 Hz. Trunk and pelvis catch and finish positions were computed to determine the range  
135 of motion (°) of these two body segments. A positive segment angle represented an extension  
136 with respect to the vertical axis.

137 The electromyographic signal of each muscle was filtered (Butterworth 4<sup>th</sup> order bandpass  
138 filter, 30-450 Hz) and quantified using the root mean square (RMS) value (mV). A 25 ms  
139 electromechanical delay was considered (Le Mansec et al., 2019). The EMG signals for the  
140 self-chosen rate condition were low pass filtered (6 Hz) to create an envelope. Then, the EMG  
141 RMS was normalised against its EMG envelop peak value during the self-chosen rate  
142 condition. Left and right RMS normalised values were averaged. In addition, trunk



143 neuromuscular control was further analysed using Directed Co-Contraction Ratio (DCCR)  
144 and the Co-Contraction Index (CCI). DCCR (1) was used to represent the balance between  
145 trunk agonists and antagonists' muscles activation (Donnelly et al., 2015), while CCI (2) gave  
146 an indication about the amount of trunk co-activation (Rudolph et al., 2000). These  
147 neuromuscular parameters were computed through the drive phase as follows:

148 If agonists' mean activation was higher than antagonists' mean activation then:

$$149 \quad DCCR = 1 - \frac{\overline{EMG}_{antagonists}}{\overline{EMG}_{agonists}} \quad \text{Else: } DCCR = \frac{\overline{EMG}_{agonists}}{\overline{EMG}_{antagonists}} - 1 \quad (1)$$

150 where the ES were defined as agonists to the movement, and RA and EO were defined as  
151 antagonists. For instance, greater ES

$$CCI = \frac{\overline{EMG}_{Lower}}{\overline{EMG}_{Higher}} * (\overline{EMG}_{Lower} + \overline{EMG}_{Higher}) \quad \text{activations induced larger DCCR.}$$

$$153 \quad (2)$$

154 where the agonists and antagonists were defined as higher or lower depending on their  
155 relative level of activity. For instance, balanced activations between extensor and flexor  
156 muscles, combined with greater level of activity induced larger CCI. But either unbalanced  
157 activation or lower level of activity decreased the CCI.

158

159 All data were analysed over the drive phase, determined by the handle position. The  
160 beginning of this phase, the catch, was defined by the minimum handle position, and the  
161 finish by the maximum handle position. Data were analysed using MATLAB software  
162 (version 9.2, MathWorks, Inc., Natick, MA, USA).

163

## 164 ***Statistics***

165 All power, kinematic and core stability variables were averaged over the middle 12 driving  
166 phases for each participant. One subject kinematic data was not usable for analysis, probably

167 because of sweating unsticking the trunk IMU, therefore the kinematic analysis was only  
168 conducted on the remaining 23 subjects.

169 Data normality was checked by the Shapiro Wilk test, and the alpha level was set at 0.05  
170 (Statistica v10, StatSoft Inc, Tusla, USA). Repeated-measures ANOVAs were used to  
171 quantify the effect of stroke rate on the mean power, T2P, M2P, power contribution of each  
172 segment, and on pelvis and trunk ROM and angles at catch and finish positions, as well as  
173 EMG RMS. Mauchly's test of sphericity was used to check for equality of variance between  
174 the three different stroke rates. If the Greenhouse–Geisser epsilon was  $> 0.75$  the Huynh Feldt  
175 correction was used, else the Greenhouse Geisser correction value was used (Girden, 1992).  
176 The Tuckey post-hoc test was used to test the direction of the effect following the ANOVA  
177 and Cohen's d test was then applied to determine the effect sizes between pairs of  
178 significantly different variables (0.2 – 0.5 interval for a small effect, 0.5 – 0.8 for a medium  
179 effect and over 0.8 for a large effect).

180 Secondly, differences in segments power, kinematics, as well as muscles activations and co-  
181 activations, between stroke rates were examined over the drive phase using one-dimensional  
182 statistical parametric mapping (SPM) using Random Field Theory (Pataky et al., 2013). SPM  
183 repeated measures ANOVA were conducted in MATLAB using the open source spm1d code  
184 (spm1d version M.0.4.8; spm1d.org), and then paired t-test SPM were computed on  
185 significant variables. After the Bonferroni correction, the alpha significance level for SPM t-  
186 tests was set at 0.017.

187

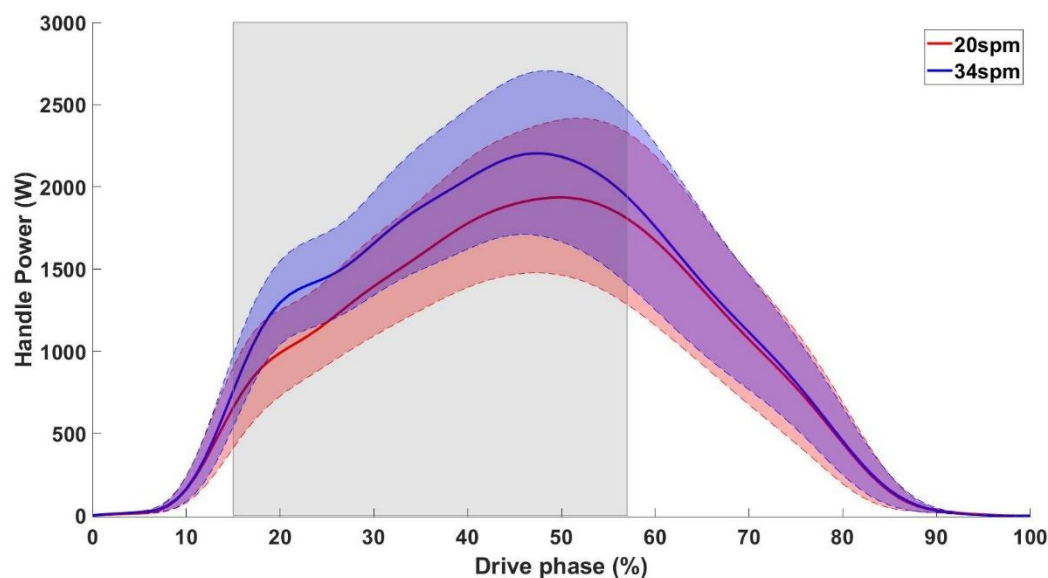
## 188 **Results**

189 The stroke rates for the three trials were 20.7 spm ( $\pm 0.5$ ), 28.0 spm ( $\pm 0.5$ ), 34.0 spm ( $\pm 0.8$ ).

190 The stroke length was not different between the conditions ( $1.67 \pm 0.12$  m for 20 spm,  
191  $1.68 \pm 0.11$  m for 28 spm,  $1.67 \pm 0.11$  m for 34 spm).

## 192 ***Performance***

193 Mean power handle significantly increased with stroke rate, from 939 W ( $\pm$  236) for 20 spm  
194 to 1012 W ( $\pm$  237) and 1060 W ( $\pm$  240) for 28 spm and 34 spm, respectively (Cohen's  $d > 0.8$   
195 for the three comparisons). SPM ANOVA indicated a larger handle power production with  
196 the increase of the stroke rate, from 15% to 60% of the drive phase ( $p < 0.05$ ;  $F = 7.421$ ). The  
197 power produced at the handle was significantly greater for 28 spm than 20 spm, but only  
198 between 16% and 57% of the drive phase. For 34 spm, the power at the handle was larger  
199 than 20 spm in the 15-57% window (Figure 2) and greater than 28 spm in the 16-54%  
200 window.



201

202 Figure 2. Group average (mean  $\pm$  SD) power production at the handle (W) during the drive  
203 phase (%) for the 20 spm (in red) and 34 spm (in blue) conditions. The grey shaded area  
204 represents the significant differences time-window ( $p < 0.017$ ).  
205

## 206 *Rowing technique*

207 The mean powers of each segment were significantly greater with stroke rate increase (Table  
208 1;  $p < 0.001$ ), with medium to large effect sizes. The segments power contribution was  
209 slightly but significantly different with the stroke rate, with a small increase in the legs power  
210 contribution at 34spm, while the one at the arms level was lower (Table 1). The  
211 corresponding effect size was medium ( $0.5 < d < 0.8$ ).

212 Table 1. Comparison of the mean  $\pm$  SD power (W) and contribution  $\pm$  SD (%) at each segment  
 213 level. Main cadence effect is reported with the p value level or NS (not significant).  
 214 \*Significant difference from 20 spm ( $p < 0.05$ ). <sup>§</sup>Significant difference from 28 spm ( $p <$   
 215 0.05).

|        | Mean power handle<br>(W)<br>$p < 0.001$ | Mean power legs<br>(W)<br>$p < 0.001$ | Legs' contribution<br>(%)<br>$p < 0.05$ | Mean power trunk<br>(W)<br>$p < 0.001$ | Trunk's contribution<br>(%)<br><i>NS</i> | Mean power arms<br>(W)<br>$p < 0.001$ | Arms' contribution<br>(%)<br>$p < 0.001$ |
|--------|-----------------------------------------|---------------------------------------|-----------------------------------------|----------------------------------------|------------------------------------------|---------------------------------------|------------------------------------------|
| 20 spm | 939 $\pm$ 236                           | 410 $\pm$ 105                         | 43.1 $\pm$ 3.8                          | 335 $\pm$ 106                          | 34.6 $\pm$ 4.5                           | 210 $\pm$ 61                          | 22.3 $\pm$ 4.5                           |
| 28 spm | 1012 $\pm$ 237*                         | 447 $\pm$ 105*                        | 43.6 $\pm$ 3.8                          | 362 $\pm$ 111*                         | 34.7 $\pm$ 4.3                           | 221 $\pm$ 60*                         | 21.7 $\pm$ 4.3*                          |
| 34 spm | 1060 $\pm$ 240* <sup>§</sup>            | 471 $\pm$ 105* <sup>§</sup>           | 43.8 $\pm$ 3.2*                         | 379 $\pm$ 111* <sup>§</sup>            | 34.6 $\pm$ 3.9                           | 229 $\pm$ 58* <sup>§</sup>            | 21.5 $\pm$ 4.1*                          |

216  
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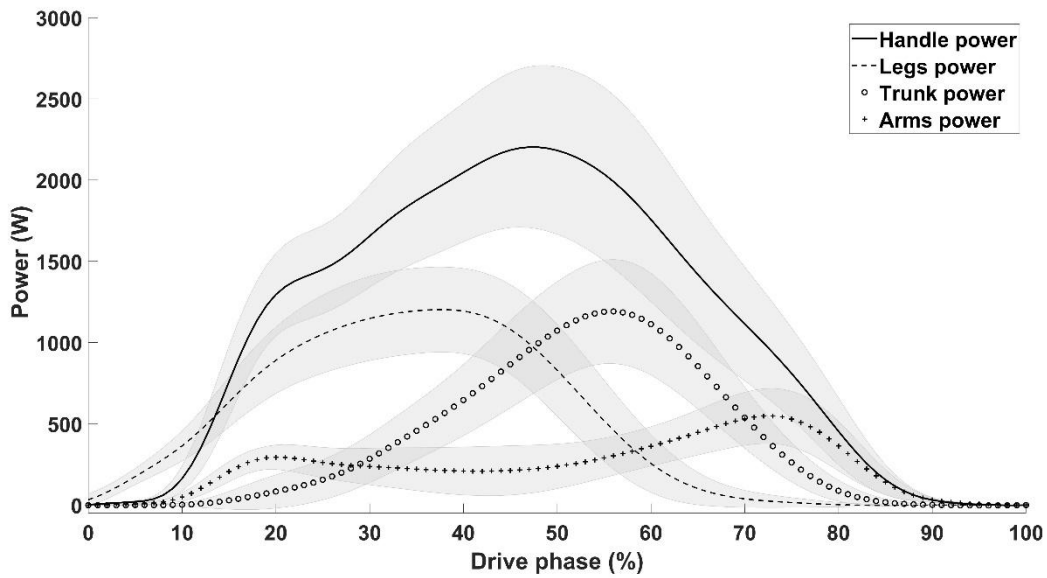
218 Rowing technique variables for the handle, the legs, the trunk and the arms are presented in

219 Table 2. Peak power of the legs occurred earlier ( $p < 0.01$ , medium effect size), while the one  
 220 for the arms was delayed ( $p < 0.001$ , small effect size) with the increase of the stroke rate. But  
 221 the peak power of the handle and the trunk remained unchanged. Moreover, mean to peak  
 222 power values of the different segments were larger for the higher stroke rate (medium to large  
 223 effect sizes for the legs and trunk, and large effect size for the arms). The corresponding  
 224 rowing style, shown in Figure 3, was close to the Rosenberg style, as defined by Kleshnev  
 225 (2016), with sequencing leg and trunk power production and an important contribution of the  
 226 trunk power to performance. The power curves and statistical comparison between 20 spm  
 227 and 34 spm are presented in Figure 4 for the legs, the trunk and the arms. Significant  
 228 differences in power (SPM ANOVA) were between the beginning and 47% for the legs,  
 229 within the 27-58% window for the trunk and the 15-34% window for the arms. For the legs,  
 230 the significant differences time-window were 2-45% between 20 spm and 28 spm and 1-36%  
 231 between the 28 spm and 34 spm. At the trunk level, differences occurred within 32-57% and  
 232 31-49% of the drive phase when comparing 20 spm and 28 spm and 28 spm and 34 spm,  
 233 respectively. Lastly, the arms power significant differences time-window was 17-29%  
 234 between 28 spm and both other rates.

235 Table 2. Comparison of mean  $\pm$  SD time to peak power (T2P, % of the drive phase) and mean  
 236 to peak power (M2P) of each segment. Main cadence effect is reported with the p value level  
 237 or NS (not significant). \*Significant difference from 20 spm ( $p < 0.05$ ). <sup>§</sup>Significant  
 238 difference from 28 spm ( $p < 0.05$ ).

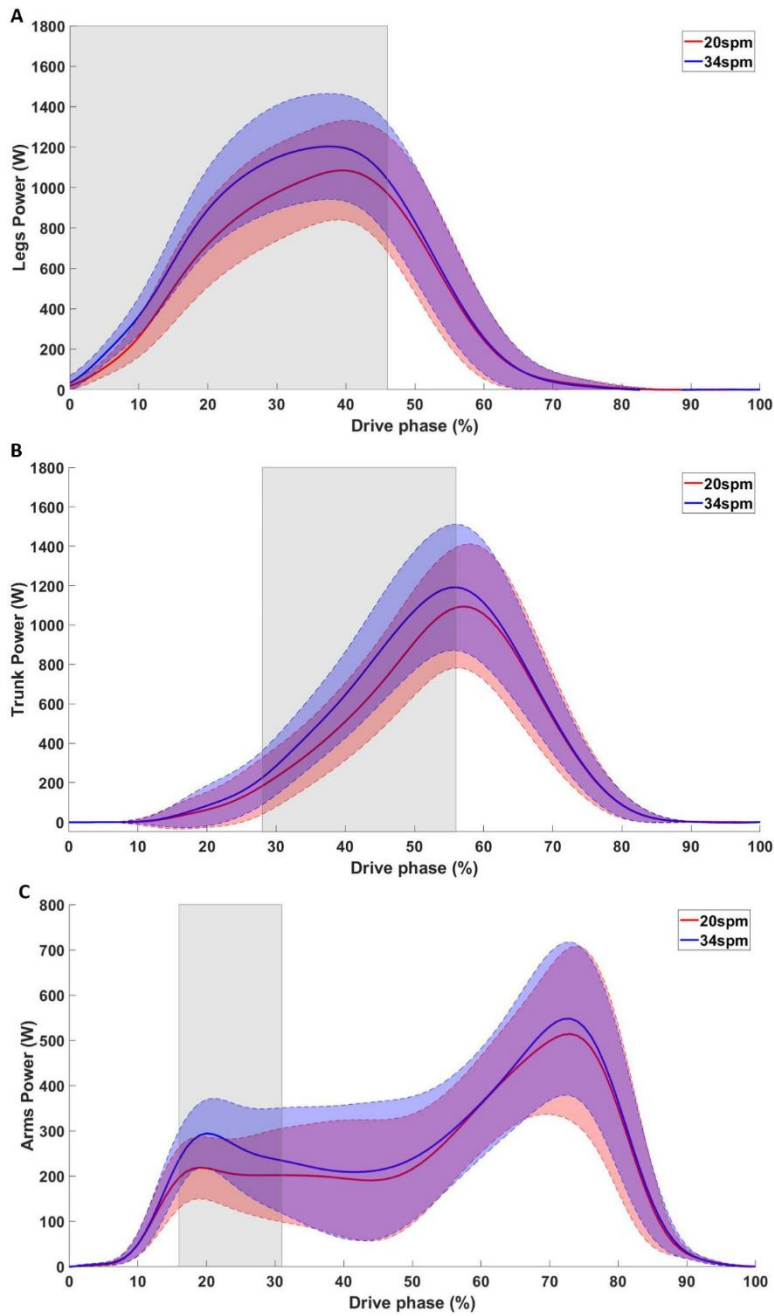
239  
240

|               | T2P Handle (%) | T2P Plegs (%)      | T2P Ptrunk (%) | T2P Parmis (%)           | M2P Handle  | M2P Plegs                 | M2P Ptrunk                | M2P Parmis                |
|---------------|----------------|--------------------|----------------|--------------------------|-------------|---------------------------|---------------------------|---------------------------|
|               | <i>NS</i>      | <i>p &lt; 0.01</i> | <i>NS</i>      | <i>p &lt; 0.001</i>      | <i>NS</i>   | <i>p &lt; 0.05</i>        | <i>p &lt; 0.001</i>       | <i>p &lt; 0.001</i>       |
| <b>20 spm</b> | 53.1 ± 3.9     | 43.2 ± 3.4         | 60.4 ± 3.0     | 75.4 ± 3.6               | 0.48 ± 0.03 | 0.37 ± 0.03               | 0.29 ± 0.03               | 0.39 ± 0.08               |
| <b>28 spm</b> | 52.1 ± 3.1     | 41.5 ± 4.2*        | 60.0 ± 2.5     | 75.9 ± 2.9*              | 0.47 ± 0.02 | 0.37 ± 0.03               | 0.30 ± 0.03*              | 0.41 ± 0.08*              |
| <b>34 spm</b> | 52.1 ± 2.5     | 41.2 ± 4.5*        | 59.9 ± 2.2     | 76.5 ± 2.3* <sup>§</sup> | 0.48 ± 0.02 | 0.38 ± 0.04* <sup>§</sup> | 0.31 ± 0.03* <sup>§</sup> | 0.42 ± 0.09* <sup>§</sup> |



241  
242  
243  
244

Figure 3. Group average power (W) production at the handle and for each segment, during the drive phase (%) for the 34 spm condition. All three stroke rates visually depicted similar patterns.



245  
 246 Figure 4. Group average (mean  $\pm$  SD) power production of the legs (A), trunk (B) and arms (C) power  
 247 production (W) during the drive phase (%) for the 20 spm (in red) and 34 spm (in blue) conditions.  
 248 The  
 249 grey shaded area represents the significant differences time-window ( $p < 0.017$ ).  
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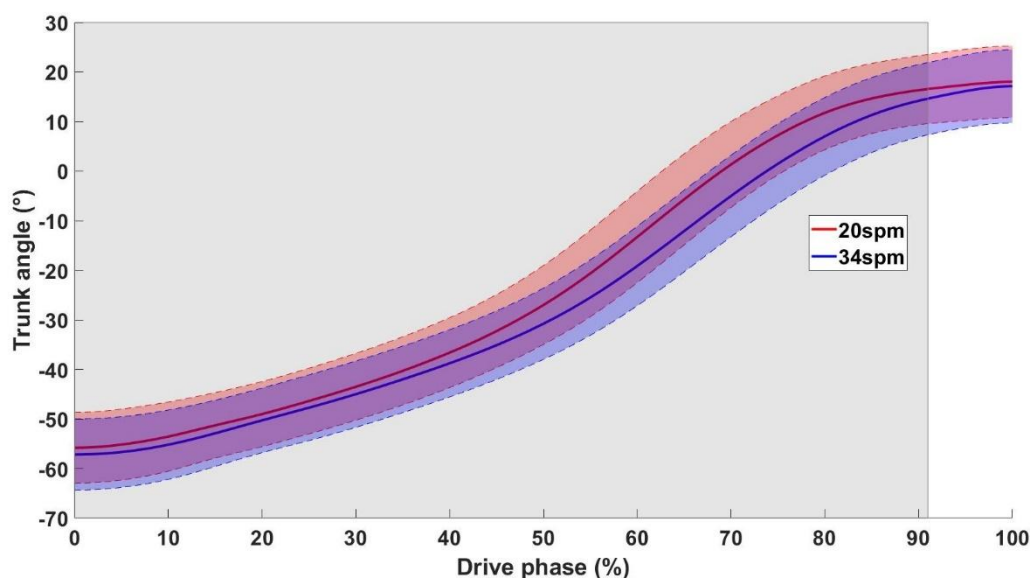
251 **Core Stability kinematics**

252 Only the trunk angle at the catch position was significantly influenced by the stroke rate, with  
 253 a larger forward lean (with a medium effect size) at the higher stroke rate (Table 3). Pelvis  
 254 angle profiles through the drive phase demonstrated a retroversion but did not differ between  
 255 the conditions. However, trunk angle profiles differed significantly between all rates (SPM  
 256 ANOVA  $p < 0.05$ , in the 0-93% window). The significant differences between 20 spm and 34  
 257 spm occurred in the 0-91% window (Figure 5). Differences were also significant between the  
 258 other rates, but in a lesser extent, i.e. within 49-76% between 20 spm and 28 spm, and within  
 259 58-93% between 28 spm and 34 spm. The trunk remained in a flexed position from the  
 260 beginning to 70-80% of the drive phase and extended then till the end of the drive phase. The  
 261 higher stroke rate induced a delayed trunk extension starting roughly at 50% of the drive  
 262 phase, but without impacting the trunk angle at the finish position.

263 Table 3. Comparison of mean  $\pm$  SD ranges of motion (ROM,  $^{\circ}$ ) and angles ( $^{\circ}$ ) of each  
 264 segment. Main cadence effect is reported with the p value level or NS (not significant).  
 265 \*Significant difference from 20 spm ( $p < 0.05$ ).

|        | Pelvis ROM ( $^{\circ}$ ) | Trunk ROM ( $^{\circ}$ ) | Pelvis catch position ( $^{\circ}$ ) | Trunk catch position ( $^{\circ}$ ) | Pelvis finish position ( $^{\circ}$ ) | Trunk finish position ( $^{\circ}$ ) |
|--------|---------------------------|--------------------------|--------------------------------------|-------------------------------------|---------------------------------------|--------------------------------------|
|        | NS                        | NS                       | NS                                   | $p < 0.001$                         | NS                                    | NS                                   |
| 20 spm | 42.4 $\pm$ 7.5            | 74.9 $\pm$ 8.8           | -0.4 $\pm$ 9.9                       | -55.8 $\pm$ 7.1                     | 40.5 $\pm$ 12.0                       | 18.0 $\pm$ 7.2                       |
| 28 spm | 42.5 $\pm$ 8.1            | 75.8 $\pm$ 8.1           | 0.1 $\pm$ 10.2                       | -56.6 $\pm$ 7.2*                    | 41.2 $\pm$ 11.3                       | 18.4 $\pm$ 7.2                       |
| 34 spm | 42.4 $\pm$ 7.2            | 74.9 $\pm$ 6.5           | 0.7 $\pm$ 10.4                       | -57.2 $\pm$ 7.2*                    | 41.5 $\pm$ 10.9                       | 17.2 $\pm$ 7.4                       |

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269 Figure 5. Group average (mean  $\pm$  SD) trunk angle ( $^{\circ}$ ) during the drive phase (%) for the 20  
 270 spm (in red) and 34 spm (in blue) conditions. The grey shaded area represents the significant  
 271 differences time-window ( $p < 0.017$ ).

272

273 ***Core Stability neuromuscular control***

274 All muscles had a greater activation with the stroke rate increase, especially at 34 spm, except  
 275 the ES (Table 4 and Figure 6). The effect sizes for RA were medium, and medium to large for  
 276 EO, TF and GM muscles. This was true whatever the variable considered (EMG  
 277 quantification or EMG profile) over the whole drive phase (SPM ANOVA  $p$ -values  $< 0.05$ ). In  
 278 addition, RA were more activated during 60% to 100% of the drive phase for the 28 spm  
 279 condition compared to 20 spm, and during 64% to 87% for the 34 spm condition with respect  
 280 to 28 spm. For EO, the significant time-window where differences occurred was between  
 281 78% and 100% when comparing 20 spm and 28 spm. In the 34 spm condition, a larger  
 282 activation was found within 75-100% of the drive phase compared to 28 spm. TF activation  
 283 differences started at 59% and 64% until the end of the drive phase, for comparison between  
 284 20 spm and 28 spm and between 28 spm and 34 spm, respectively. For GM, a significant  
 285 larger activation was noticed in the early phase (0-16%) for 28 spm compared to 20 spm,  
 286 whereas no difference was noted between 28 spm and 34 spm.

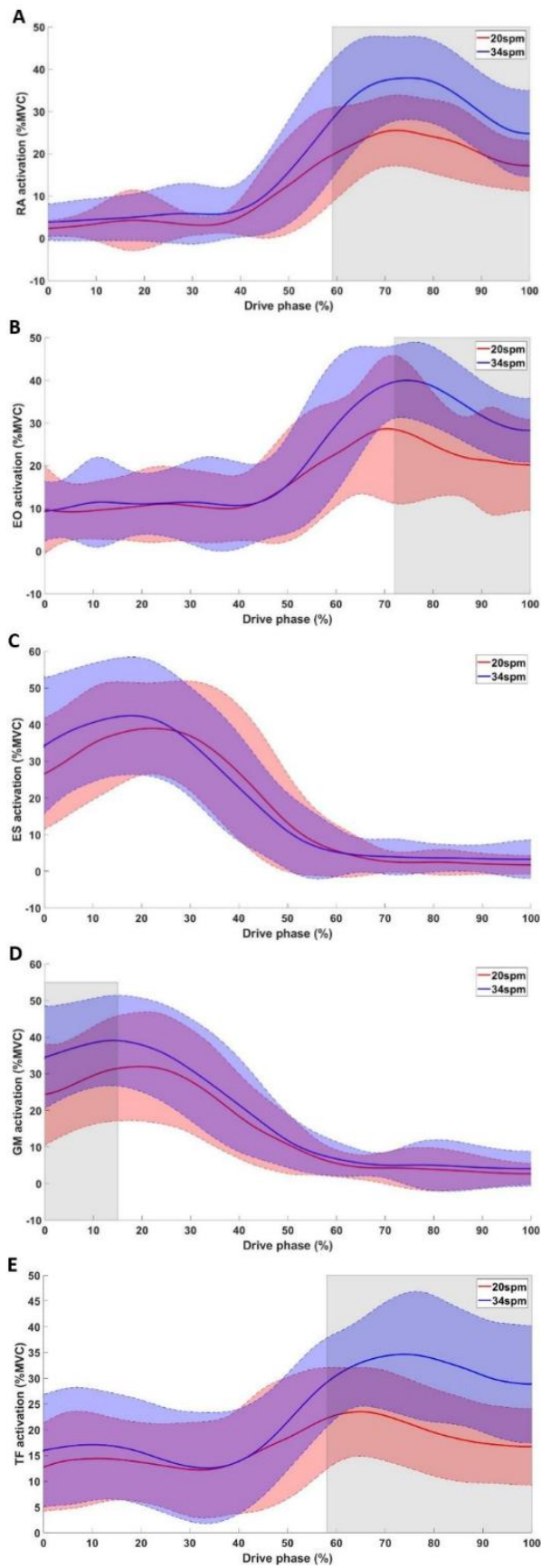
287 Table 4. Comparison of mean  $\pm$  SD of EMG activations (%MVC) of the rectus abdominis  
 288 (RA) muscles, the external oblique (OE) muscles, the erector spinae (ES) muscles, the tensor  
 289 of the fascia lata (TF) and the gluteus maximus (GM) muscles. Main cadence effect is  
 290 reported with the  $p$  value level or NS (not significant). \*Significant difference from 20 spm ( $p$   
 291  $< 0.05$ ). <sup>s</sup>Significant difference from 28 spm ( $p < 0.05$ ).

|               | RA (%MVC)                     | OE (%MVC)                    | ES (%MVC)      | TF (%MVC)       | GM (%MVC)                    |
|---------------|-------------------------------|------------------------------|----------------|-----------------|------------------------------|
|               | $p < 0.01$                    | $p < 0.001$                  | NS             | $p < 0.01$      | $p < 0.001$                  |
| <b>20 spm</b> | 12.9 $\pm$ 3.8                | 16.8 $\pm$ 8.4               | 17.8 $\pm$ 5.5 | 14.9 $\pm$ 5.3  | 17.0 $\pm$ 5.0               |
| <b>28 spm</b> | 16.7 $\pm$ 7.1                | 18.8 $\pm$ 6.1*              | 19.1 $\pm$ 5.3 | 17.4 $\pm$ 4.1* | 20.1 $\pm$ 5.4*              |
| <b>34 spm</b> | 21.1 $\pm$ 13.2* <sup>s</sup> | 21.7 $\pm$ 7.2* <sup>s</sup> | 18.9 $\pm$ 5.4 | 18.4 $\pm$ 4.5* | 23.2 $\pm$ 5.3* <sup>s</sup> |

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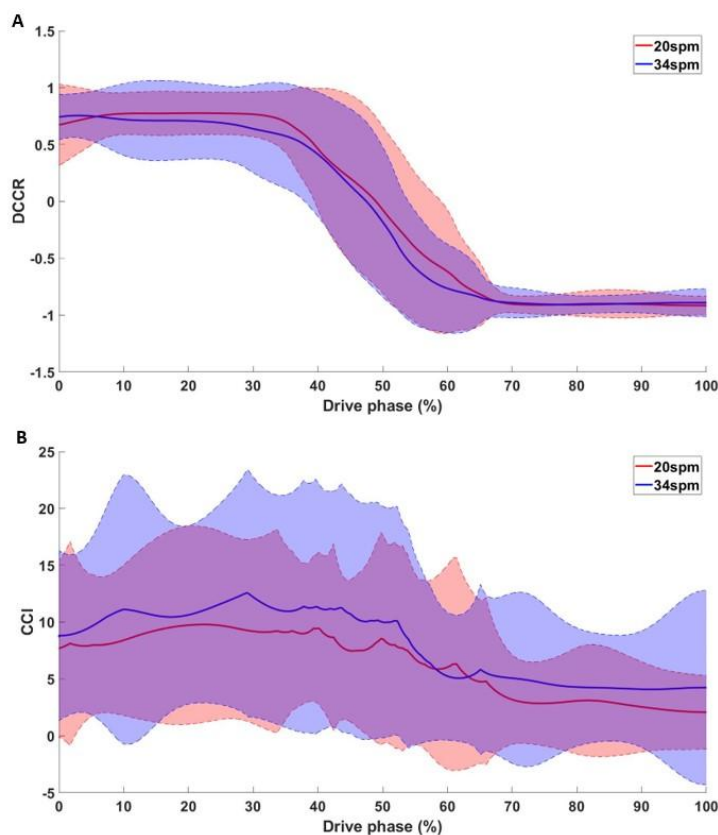


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Figure 6. Group average (mean  $\pm$  SD) RA (A), EO (B), ES (C), TF (D) and GM (E) activations (%MVC) during the drive phase (%) for the 20 spm (in red) and 34 spm (in blue)

297 conditions. The grey shaded area represents the significant differences time-window ( $p <$   
298  $0.017$ ).

299 No difference in DCCR or CCI was observed. Their patterns are presented in Figure 7. The  
300 DCCR presented a larger activation of the trunk extensor muscles than the abdominal muscles  
301 within the first 35% of the drive phase. Then, the decrease in this value indicated a transition  
302 towards a more abdominal orientated ratio, these muscles becoming the most activated for the  
303 remaining 30% of the drive phase. The CCI remained relatively low and stable across the  
304 drive phase. For example, if both extensor and flexor groups would have activated in average  
305 at 40% of their MVC, then the CCI value would have been 80. The CCI value around 10 in  
306 the first and last part of the drive phase are due to the large imbalance between the agonists  
307 and the antagonists, despite a large amount of muscles activation. During the central part of  
308 the drive phase, the more balanced activation between these muscles is impaired by a rather  
309 small amount of activation, maintaining the CCI value low.



310  
311 Figure 7. Group average (mean  $\pm$  SD) Directed Co-Contraction Ratio (A) and Co-Contraction  
312 Index (B) during the drive phase (%) for the 20 spm (in red) and 34 spm (in blue) conditions.

## 313 **Discussion and Implications**

314 The present study's aims were to determine the effect of stroke rate on rowing ergometer  
315 performance, technique and core stability. The main findings were that i) the power at the  
316 handle and at each segment level are enhanced at a higher stroke rate, mostly during the first  
317 half of the drive phase, indicating that rowing technique is modified, ii) trunk extension was  
318 delayed during the drive phase, without altering its range of motion, and iii) some core  
319 muscles showed a greater activation without any change in the co-activation along with the  
320 stroke rate increase.

321

### 322 ***Rowing performance and technique***

323 Mean powers of handle and segments were largely impacted by the stroke rate variation,  
324 showing greater power at higher stroke rate as previously reported in the literature at the  
325 handle level (Buckeridge et al., 2012; Held et al., 2020; Hofmijster et al., 2007; Kleshnev,  
326 1996; McGregor et al., 2004), legs level (Held et al., 2020) and every segment level  
327 (Kleshnev, 1996). The relative contributions of each segment to the total power (in %) were  
328 close to those reported by Kleshnev (2000). Although these contributions were influenced by  
329 the stroke rate, the variation remained below 1% and should not be considered as a substantial  
330 modification of the segments contribution. The greater handle power at higher stroke rate  
331 occurred roughly between 15% and 55% of the drive phase, and is likely to be explained by  
332 the increase in legs, trunk and arm power. Indeed, legs and trunk powers were impacted by  
333 the stroke rate during the largest part of their power production until their peak power  
334 delivery. Consequently, legs time to peak power occurred earlier, while the trunk time to peak  
335 power remained unchanged. Although these timing results were different from Kleshnev  
336 (1996), one should keep in mind that these power profiles, and hence their sensitivity to  
337 stroke rate manipulation, are strongly related to the technical styles, depending on the rower's

338 level (Buckeridge et al., 2012; Cerne et al., 2013) and probably on training background (Hill,  
339 2002; Kleshnev, 2016; Warmenhoven et al., 2018B). Surprisingly, although differences in  
340 arms power were also observed at a higher stroke rate, these took place only at the beginning  
341 of the drive phase too, where only 50% of their maximal power is produced. But arms time to  
342 peak power was not in line with this power profile alteration, as it occurred later with the  
343 stroke rate increase. This shows the importance of this early phase of the drive to enhance  
344 performance through a faster rowing movement. During this phase, rowers might have to  
345 provide more power at their arms level to tense the ergometer chain following a faster start at  
346 a higher stroke rate. The lack of power profiles difference in the second half of the drive  
347 phase, especially at the arms level would explain the comparable handle power production  
348 during the last 40%, despite a higher stroke rate. This power alteration resulted in a greater  
349 mean to peak power ratio at each level, showing a greater average power relative to the peak  
350 power at higher stroke rate. These more "rectangular" or "flattened" curves could be related to  
351 greater power output, as observed for elite vs. sub-elite rowers (Smith & Draper, 2006).  
352 Therefore, in order to improve their performance at each stroke rate, it is advised to produce a  
353 high power at each segment level over the longest portion possible over the drive phase.  
354 Moreover, given that stroke rate influences the distribution between the drive and the  
355 recovery times, it is recommended to analyze technical and biomechanical variables, in  
356 particular time-dependent variables (such as time to peak), over the drive phase only.  
357 Furthermore, the rowing technical style, defined in Figure 3, remained unchanged with the  
358 stroke rate and close to the Rosenberg style (Kleshnev, 2016). The results presented in this  
359 study might not be generalizable to other rowing styles, so future research should consider  
360 this in order to better understand the stroke rate effect with respect to the rowing style. Other  
361 factors could modify the effect of the stroke rate on the rowing technique as sex  
362 (Attenborough et al., 2012; Ng et al., 2013), boat-side (Warmenhoven et al., 2018A), or

363 expertise (Buckeridge et al., 2012). Therefore, larger sample size would be needed to  
364 investigate further these potential covariates effects.

365

### 366 *Core Stability*

367 The significant contribution of the trunk to power production with this technical style, i.e.  
368 over 30%, highlights the importance of a core stability analysis to better understand its  
369 adaptation to the stroke rate. Contrarily to our hypothesis, the trunk and pelvis ranges of  
370 motion were not larger for the higher stroke rate, whereas Li et al. (2020) found that the  
371 thoracic range of motion was increased between a 18 spm and a 32 spm rowing condition  
372 (e.g. by 3.9°). Moreover, the average trunk range of motion in their study, whatever the  
373 condition, was around 40°, while ours rather reached 75°. These differences may be the  
374 consequence of a different rowing style, and/or the trunk kinematic model. Indeed, the IMU  
375 sensor position at the second thoracic vertebrae level in Li et al.'s study (2020), could have  
376 been sensitive to a forward roll of the shoulder during rowing, thus reducing the trunk range  
377 of motion. In addition, the main difference between our setups was the instruction given to the  
378 rowers, as these authors asked for a long training session intensity (i.e. relatively low),  
379 whereas rowers had to perform their strokes at maximal intensity in the current study. As this  
380 rowing intensity has an impact on lower limb joint powers (Ettema et al., 2022), it is likely  
381 that this could influence trunk kinetics and kinematics. These different points could explain  
382 the different effect of stroke rate on trunk kinematics between the present investigation and Li  
383 et al.'s study (2020). As a large trunk range of motion (with respect to the pelvis) has been  
384 linked to better rowing performance (Buckeridge et al., 2016A), we could hypothesise that the  
385 large values measured in the present study underline that the trunk range of motion already  
386 levelled off to produce the highest performance, given the maximal intensity instruction, and  
387 regardless of the stroke rate. The same suggestion could be made for the pelvis kinematics, as

388 it remained at the same level, contrarily to previous studies (McGregor et al., 2004; Li et al.,  
389 2020). However, the aforementioned studies used a fixed rowing ergometer, which enables  
390 pelvic kinematics alterations with the stroke rate, whereas free-floating ergometers don't  
391 (Hislop et al., 2010). Interestingly, the gluteus maximus had a greater activation in the early  
392 part of the drive phase, while the pelvis kinematics didn't differ. Thus, as the stroke rate rises,  
393 their role might be to increase the leg power production by extending the hip, rather than to  
394 support further the pelvis retroversion.

395 Despite the absence of difference in the trunk range of motion, we observed a delayed  
396 extension of the trunk at the higher stroke rate. This delay occurred mostly from 50% to 90%  
397 of the drive phase, so after the trunk power significant increase, which ended up around 55%.  
398 As erector spinae muscles did not present a larger activation and trunk extension was delayed,  
399 this indicates that the larger trunk power would not be the consequence of a trunk-located  
400 mechanism of power production, but probably because of a transfer from the lower limb  
401 kinetics. Therefore, when the stroke rate rises, it is possible that the trunk might act more as a  
402 force transferrer than a force producer. Other mechanisms should be investigated to enhance  
403 our understanding of this force transfer, as for instance the role of passive structures (e.g.  
404 thoracolumbar fasciae), intra-abdominal pressure or deeper core muscles (e.g. transversus  
405 abdominis). Besides, given that the trunk remained in a flexed position during this force  
406 transfer phase, the erector spinae activation level measured in this study might have already  
407 been sufficient to support this force transfer, whatever the stroke rate.

408 Also, no trunk co-activation strategy has been noticed in the present study, neither with  
409 DCCR nor with CCI. Indeed, as depicted in the Figure 7, a clear transition between erector  
410 spinae activations and abdominal muscles activations in the middle of the drive phase,  
411 combined to the absence of any CCI rise during the 40-60% window of the drive phase,  
412 revealed that the co-activation measurements could not help the understanding of force

413 transfer mechanisms through the core during ergometer rowing. A high-level of co-activations  
414 between abdominals and spine erectors is therefore not required, whatever the stroke rate.

415 The statistical parametric mapping analysis also underlined the larger abdominal activations  
416 in the late part of the drive phase, that acted eccentrically to control the trunk delayed  
417 extension (Pollock et al., 2009). Simultaneously, the larger tensor fasciae latae activation  
418 probably helped maintaining pelvic retroversion's stability. All these muscles activities also  
419 enabled to anticipate the recovery phase where a trunk flexion and a pelvic anteversion would  
420 be required. With the increased stroke rate, core muscles larger activations might reduce the  
421 recovery phase duration to perform the demanded stroke rate (Li et al., 2020). As more  
422 abdominal and tensor fasciae latae forces might be necessary as the stroke rate rises, we could  
423 therefore advice to target eccentric abdominal training for rowers.

424

#### 425 *Summarizing conclusion*

426 Our results show that ergometer rowing at a higher stroke rate induces greater power  
427 production at the handle and at each body segment during the first half of the drive phase, as  
428 well as adaptations in the rowing technique.

429 Stroke rate also influences core stability, as illustrated by the delayed trunk extension,  
430 controlled by the eccentric action of the abdominal muscles. The core might further act as a  
431 force transferrer with the higher stroke rate, despite any increase in the co-activation strategy.

432 Rowing at a higher stroke rate certainly places the rower closer to the technical and  
433 neuromuscular conditions encountered during a competition, but rowing at a lower stroke rate  
434 could be helpful to work on an earlier trunk extension technique, while maintaining erectors  
435 spinae levels of activation. Moreover, to improve the force transfer capacity through the core,  
436 training at higher stroke rate should be more efficient. Further work is needed to better  
437 understand force transfer mechanisms through the trunk during rowing tasks.

438 ***Acknowledgement***

439 We thank the participants and the Pole France et Espoir d’Aviron de Nancy for their  
440 involvement and time. The authors are also grateful to Arnaud Cosson, CEO of HRV  
441 Simulation, for supporting F. Simon during his PhD thesis.

442 ***Disclosure statement***

443 No potential conflict of interest was reported by the author(s).

444

445 ***Funding***

446 This work was supported by the INSEP under Grant [project 19r22] and FEDER FSE-IEJ  
447 Lorraine et Massif des Vosges 2014-2020 under Grant [LO0027294]

448



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