

Center of pressure displacement due to graded controlled perturbations to the trunk in standing subjects:
the force-impulse paradigm

Original

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(Article begins on next page)

1
2 *Original Article*
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4 **Center of pressure displacement due to graded controlled**
5 **perturbations to the trunk in standing subjects: the force-**
6 **impulse paradigm**
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24 **ABSTRACT**

25 *Purpose:* Many studies have investigated postural reactions (PR) to body-delivered
26 perturbations. However, attention has been focused on the descriptive variables of
27 the PR rather than on the characterization of the perturbation. This study aimed to
28 test the hypothesis that the impulse rather than the force magnitude of the
29 perturbation mostly affects the PR in terms of displacement of the center of foot
30 pressure (ΔCoP).

31 *Methods:* Fourteen healthy young adults (7 males and 7 females) received two
32 series of 20 perturbations, delivered to the back in the anterior direction, at mid-
33 scapular level, while standing on a force platform. In one series, the perturbations
34 had the same force magnitude (40 N) but different impulse (range: 2-10 Ns). In the
35 other series the perturbations had the same impulse (5 Ns) but different force
36 magnitude (20-100 N). A simple model of postural control restricted to the sagittal
37 plane was also developed.

38 *Results:* The results showed that ΔCoP and impulse were highly correlated (on
39 average: $r=0.96$) while the correlation ΔCoP -force magnitude was poor ($r=0.48$)
40 and not statistically significant in most subjects. The normalized response,
41 $\Delta\text{CoP}_n=\Delta\text{CoP}/I$, was independent of the perturbation magnitude in a wide range of
42 force amplitude and impulse and exhibited good repeatability across different sets
43 of stimuli (on average: $\text{ICC}=0.88$). These results were confirmed by simulations.

44 *Conclusion:* The present findings support the concept that the magnitude of the
45 applied force alone is a poor descriptor of trunk-delivered perturbations and suggest
46 that the impulse should be considered instead.

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47 **Keywords:** Postural reaction; perturbation; force; impulse; center of pressure; balance
48 control.

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50 1. INTRODUCTION

51 Research on postural reactions (PR) has employed a variety of perturbation
52 techniques intended to simulate in laboratory conditions the events that challenge
53 the body equilibrium in real life. Two distinct approaches have been followed:
54 imparting the perturbation i) to the base of support by sliding or tilting the platform
55 (Schmidt et al. 2015; Grassi et al. 2017; Robbins et al. 2017) or ii) directly to the
56 upper body. These two perturbation modes elicit fundamentally different PR
57 (Bortolami et al. 2003; Colebatch et al. 2016; Chen et al. 2017) and thus are both
58 worth to be pursued. However, while the moving platform is easily described and
59 standardized in terms of extent and speed of displacement and rotation, description
60 and quantification of upper body perturbation are more difficult. Direct body
61 perturbation has been achieved in the most disparate of ways. Some devices were
62 based on imparting a pull force to the body by the sudden release of a weight
63 connected to the body via a cable (Martinelli et al. 2015; Maaswinkel et al. 2016;
64 Azzi et al. 2017) or employing electric actuators (Pidcoe and Rogers 1998;
65 Sturnieks et al. 2013; Fujimoto et al. 2015; Robert et al. 2018), which, however,
66 alter the subject's resting posture, thus potentially affecting the overall PR. Others
67 are based on the application of a push force imparted manually by pushing the
68 subject with the hands (Colebatch et al. 2016), or by releasing a pendulum which
69 hits the body at shoulder level (Kim et al. 2012), or by the action of a hand-held
70 device which records the force profile during contact with the subject (Kim et al.
71 2009; Pasman et al. 2019; Dvir et al. 2020). In most cases little attention was
72 devoted to the characterization of the perturbation and the relation between the
73 magnitude of the perturbation and the postural response, focusing instead on the

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2 74 factors affecting CoP steadiness (Martinelli et al. 2015; Azzi et al. 2017; Grassi et
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4 75 al. 2017) or its association with the risk of falling (Sturnieks et al. 2013; Fujimoto
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6 76 et al. 2015). However, the precise identification of the input variable that better
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8 77 correlates with the CoP response could facilitate the interpretation of the results and
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10 78 the design of appropriate postural tests. Significantly, it could enhance testing of
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12 79 patients affected with disorders in which the normal PR may be compromised
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14 80 (Grassi et al. 2017; Colebatch and Govender 2019).

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19 81 Although it is generally acknowledged that, within the boundaries of stability, the
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21 82 greater the magnitude of the perturbation the greater is the PR (Diener et al. 1988;
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23 83 Kim et al. 2009; Azzi et al. 2017; Forghani et al. 2017; Teixeira et al. 2019), very
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25 84 few studies investigated this relation with upper body-directed perturbation. Kim et
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27 85 al (2009) evidenced a positive correlation between the peak force of a body-directed
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29 86 push perturbation and the displacement of the center of pressure (CoP). However,
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31 87 by exploring specifically this facet of PR, we have recently observed that in young
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33 88 men, the magnitude of the CoP response, in terms of its displacement, was better
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35 89 correlated with the impulse than with the peak force of the postural perturbation
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37 90 (Dvir et al. 2020). On one hand, it may seem obvious that the magnitude of the
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39 91 perturbation cannot be simply characterized by the magnitude of the force but
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41 92 should also depend on the duration of the push. On the other hand, the impulse,
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43 93 indeed defined as the integral of force over time, has surprisingly not gained much
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45 94 consideration in the literature, even though it corresponds to the momentum
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47 95 transferred to the body. As such, it is directly related to the change in speed of the
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49 96 body and thus to the energy transmitted by the perturbation.
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2 97 The preliminary observation presented in Dvir et al.(2020) did not provide a clear-
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4 98 cut indication with regard to the impulse vs. force paradigm, possibly because of
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7 99 data dispersion. The postural perturbations were manually delivered, with high
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10 100 intra- and inter- subject variability, in terms of force amplitude, duration and
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12 101 impulse. This could have accounted for the intra-subject variability of the response
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14 102 and the low Pearson correlation coefficient values observed in some subjects.
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17 103 Aim of the present study is to reinvestigate the hypothesis that the CoP
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19 104 displacement due to trunk-directed push perturbations is linearly correlated with the
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21 105 magnitude of the impulse and not with the force magnitude, by means of a renewed
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23 106 experimental approach and model simulations. In order to reduce the variability in
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25 107 the magnitude of the perturbations a novel pneumo-tronic device was developed,
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27 108 capable of imparting simultaneous force- and duration-controlled perturbations
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29 109 (Ferraresi et al. 2020a, b; Maffiolo et al. 2020). In addition, the experimental results
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31 110 are discussed and compared with a simulation of the CoP response based on a
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33 111 simple single-link inverted pendulum model.
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41 112 **2. METHODS**

42 113 *2.1 Experimental test*

43 114 2.1.1 Subjects

44 115 A group of 14 healthy young adults, 7 females (mean(SD) age: 22.7(1.7)years;
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46 116 height: 1.62(0.05)m; weight: 54.0(4.2)kg; BMI: 20.7(1.5)kg/m²) and 7 males
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48 117 (mean(SD) age: 23.1(2.7)years; height: 1.78(0.11)m; weight: 70.3(6.0)kg; BMI:
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50 118 22.3(1.6)kg/m²), was recruited from the student population at the Politecnico di
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53 119 Torino. Exclusion criteria included: recent lower extremity injury and/or fracture
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2 120 (< 1 year), previous reconstructive surgery in the lower extremity and balance
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4 121 deficits. All subjects provided written informed consent to participate in this study
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6 122 which was approved by the institutional review board of the University of Torino
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9 123 (Prot. n. 380583).

14 124 2.1.2 Task and instrumentation

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16 125 The experimental task consisted of recovering balance following impulsive
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18 126 perturbations applied to the trunk in the anterior direction while standing on a force
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21 127 platform.

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24 128 The force platform, a modified Shekel (Beit Keshet, Israel) device, was made up of
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26 129 a still plate (52x36 cm) which was supported by 4 uniaxial load cells (TEDEA,
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28 130 Israel, model 1042, rated capacity 100 kgf), mounted on a base plate. The
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31 131 perturbation was applied by a pneumo-tronic perturbator designed and constructed
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34 132 at the Dept. of Mechanical and Aerospace Engineering at the Politecnico di Torino.
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36 133 The instrument is shown in Fig. 1A and was described in detail in another
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38 134 publication (Ferraresi et al. 2020b). The closed-loop force feedback design, based
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41 135 on the continuous monitoring of the perturbation force provided by a load cell
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43 136 positioned in series with the tip of the perturbator, allows for the regulation of the
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46 137 precise intensity and duration of the stimulus delivered to the subject, irrespective
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49 138 of the mechanical compliance of the operator (Ferraresi et al. 2020b).

53 139 2.1.3 Procedure

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55 140 During the test, the subjects stood barefoot on the force platform with the feet at
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58 141 pelvic distance and with vision unobstructed. Subjects were asked to assume a

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2 142 normal-relaxed stance and they were instructed to respond naturally. The feet
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4 143 locations were traced onto the platform surface to ensure consistent initial foot
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7 144 placement across test sessions for each participant. The operator stood behind the
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9 145 subject holding the perturbator while the interface was maintained at a distance of
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12 146 about 2 cm from the subject's back (Fig. 1B). Immediately before the starting of
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14 147 the test, participants were familiarized with the procedure by receiving few
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16 148 perturbations. The perturbations were delivered to the trunk always at inter-scapular
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18 149 level (IS), given that, at this site, more reproducible responses could be obtained,
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20 150 compared to lumbar level (Dvir et al. 2020).
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25 151 The test comprised two series, with a break of 5 min in between. In one series,
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27 152 namely the constant-force series, the perturbations had the same force magnitude
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29 153 (40 N), but different impulse values (2 Ns; 4 Ns; 6 Ns; 10 Ns). In the other series,
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31 154 namely the constant-impulse series, the perturbations had the same impulse (5 Ns)
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33 155 but different force magnitude (20 N; 40 N; 60 N; 100 N). Based on our previous
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35 156 experience, we operated in a range of values large enough to elicit a clearly
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37 157 detectable response and small enough to exclude a step response. The values of 40
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39 158 N and 5 Ns were arbitrarily chosen as intermediate values within that range. The
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41 159 average force perturbation profiles, for each condition, are shown in Fig. 2.
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47 160 In each series, the subjects received a total of 20 perturbations, 5 for each force
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49 161 profile mentioned above. The sequences of perturbations, each one including 5
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51 162 equal stimuli, were provided in random order. An inter-perturbation pause of at
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53 163 least 10 s was allowed for returning to relaxed stance. The order of the 2 series was
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55 164 randomized as well. A typical testing session lasted about 20 minutes.
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2 165 2.1.4 Data processing
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4 166 Data were extracted and processed with custom routines developed in
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6 167 MATLAB_R2019b®. The force signal was acquired at 1000 Hz and digitally low-
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9 168 pass filtered using a dual-pass 8th order Butterworth filter with a cut-off frequency
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11 169 of 200 Hz. The actual magnitude of the perturbation was characterized in terms of:

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15 170 • Force Amplitude (in N): the average force at the plateau. The start and the
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17 171 end of the plateau were automatically detected as the time instants at which
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19 172 the force signal crossed a threshold equal to 95% of the intended force
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21 173 magnitude (see Fig. 3).
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24 174 • Impulse (in Ns): the integral of force computed over the time interval in
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26 175 which the force is greater than 0.5 N.
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30 176 The ground reaction forces were acquired at 1000 Hz and were used to calculate
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32 177 the coordinates of the CoP. Both coordinates were digitally low-pass filtered with
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34 178 a dual-pass 8th order Butterworth filter with a cut-off frequency of 20 Hz. The
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36 179 postural response, ΔCoP , was computed as the maximum CoP displacement,
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38 180 observed within 2 s from the perturbation. The displacement (in cm) is calculated
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40 181 from the average resting position, calculated over the 3 s preceding the perturbation.
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47 182 2.1.5 Statistical Analysis
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49 183 All statistical procedures were conducted using MATLAB_R2019b®.
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52 184 Possible differences in impulse and force amplitude among the different
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54 185 perturbation types were analyzed through a Friedman test with grouping factor
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2 186 impulse and force amplitude for the constant-force and constant-impulse series,
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8 188 Pearson's correlation coefficient (r) was used to assess the relationship between
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10 189 Δ CoP and the perturbation. The Fisher's Z transform was used to estimate an
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12 190 average correlation coefficient over all subjects. Pearson's coefficient was also
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14 191 calculated to evaluate the relationship between the postural response and the
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16 192 physical characteristics of the subjects. The Friedman's test was used to determine
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18 193 whether the impulse or force amplitude affect the CoP displacement.

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23 194 Intraclass correlation coefficients ($ICC_{3,k}$), based on a mean rating ($k = 5$), absolute
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25 195 agreement, 2-ways mixed effects model were derived to quantify the reliability of
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27 196 the CoP response among different stimulus magnitudes while the coefficient of
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29 197 variation (CoV) was used to assess the variability of the responses to the same
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31 198 perturbation type. In order to evaluate whether general postural adjustments in
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33 199 anticipation of back perturbations took place during the test, changes in resting CoP
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35 200 were assessed within each session (comparing the beginning and the end of each
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37 201 experimental session, average CoP computed 30-s intervals with no perturbations;
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39 202 Wilcoxon Signed Rank Test) as well as within each of the 8 sequences of stimuli
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41 203 of the same type (comparing the 3-s CoP baseline preceding the first stimulus and
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43 204 the last one of the sequence; Wilcoxon Signed Rank Tests, with Bonferroni
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45 205 correction).

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50 206 Data in the text are expressed as mean \pm standard deviation.

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56 207 *2.2 Single-link inverted pendulum models*
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2 208 The human body orthostatic position perturbed with low entity disturbances
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4 209 occurring in the sagittal plane can be schematized by means of an inverse pendulum
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6 210 model (Winter et al. 1998). The basic scheme, implemented in MATLAB®
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8 211 Simulink® environment, represents the body as a rigid link having a single
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10 212 rotational degree of freedom (DoF) about the ankle joint (Fig. 4). For small
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12 213 oscillations of the body θ , the linearization of the model yields the following
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14 214 equations:

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$$\tau + mgd\theta - md^2 \frac{d^2\theta}{dt^2} - I \frac{d^2\theta}{dt^2} + F_e h_F = 0 \quad (1)$$

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23 216
$$CoP = \frac{-\tau - R_x h}{mg} \quad (2)$$

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26 217 where τ is the correcting torque at the ankle, m is the body mass, g is the
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28 218 gravitational acceleration, d is the distance between ankle joint and the center of
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30 219 mass (CoM), I is the rotational inertia of the body about the CoM, h_F is the distance
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32 220 between ankle joint and the point of application of the perturbation force F_e , CoP is
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34 221 the center of pressure position, R_x is the horizontal component of the ground
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36 222 reaction force, h is the height of ankle joint with respect to the fixed base of support.

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39 223 Although simplified models of balance control can focus on muscle stiffness alone
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41 224 as the main tool to achieve stabilization in quiet standing, it is well known that such
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43 225 passive behavior is generally not sufficient to ensure stability (Morasso et al. 1999),
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45 226 especially when significant external disturbances are considered. For this reason,
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47 227 the correcting torque at the ankle τ has been modeled as the sum of a passive and
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49 228 an active contribution. The passive contribution is related to the visco-elastic
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51 229 behavior of human tissues and is proportional to both the deformation θ and the rate
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2 230 of deformation $\dot{\theta}$ of the joint (Engelhart et al. 2015), whereas the active contribution
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5 231 depends on the neuromuscular control managed by the central nervous system and
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7 232 can be modeled as a delayed PD (Proportional-Derivative) action (Van Der Kooij
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9 233 et al. 2005). In particular, the output of the controller, i.e., the active torque at the
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11 234 ankle, is aimed at minimizing the error θ , i.e., the current angular displacement from
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13 235 the initial standing position ($\theta=0$). The information about the current angular
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15 236 displacement is fed to the controller by noisy and delayed sensory feedback. Thus,
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17 237 a constant transmission delay was introduced as the latency between the variation
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19 238 of θ and the generation of the reflex active torque (Goodworth and Peterka 2018),
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21 239 and an additive pink noise was introduced to account for the limitations of the
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23 240 sensory system (Van Der Kooij and Peterka 2011; Boonstra et al. 2013; Goodworth
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25 241 and Peterka 2018). Proportional and derivative gains of the PD control model then
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27 242 need to be identified, to match the characteristics of a given subject and to achieve
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29 243 stability. (Van Der Kooij et al. 2005; Van Der Kooij and Peterka 2011; Goodworth
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31 244 and Peterka 2018).

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34 245 With the limited aim of investigating the theoretical dependence of the CoP
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36 246 response to force and impulse of the perturbation, the model was configured as
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38 247 follows: 1) anthropometric parameters were set equal to average values computed
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40 248 over the participants to the experimental study (with reference to Fig. 4: $m = 62$ kg,
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42 249 $l = 1.70$ m, $h = 0.1$ m, $d = 0.6l$, $I = ml^2/12$, $h_F = 1.2$ m); 2) the coefficients of the
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44 250 passive response were set according to the literature (Engelhart et al. 2015); 3) the
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46 251 latency between the generation of the active torque and the variation of θ was set to
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48 252 the constant value of 90 ms, according to the literature (Goodworth and Peterka
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2 253 2018); 4) active control parameters and noise level were estimated by an iterative
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4 254 least-squares fitting used to match the simulation with the average experimental
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7 255 postural response.
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10 256 The CoP response to a given perturbation was obtained from the average of 5
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12 257 distinct simulations, thus accounting for the variability introduced by sensory noise.
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16 17 258 **3. RESULTS** 18

19 259 *3.1 Results of the experimental trials* 20

21 260 A representative recording of a single perturbation along with the postural response
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23 261 is shown in Fig. 3.
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28 262 The actual magnitudes for the different experimental perturbation types are shown
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30 263 in Fig. 5 for the two series. In the constant-force series, the perturbator delivered
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32 264 stimuli with different impulses and with similar force amplitude values (on average,
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34 265 39.54 ± 3.01 N) although the actual force amplitude appeared to depend on stimulus
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36 266 type ($p < 0.01$) (Figure 5A). Similarly, the perturbation types in the constant-
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38 267 impulse series were well characterized by distinct force values and similar impulse
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40 268 values (on average, the impulse was equal to 4.60 ± 0.28 Ns) although a significant
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42 269 dependence of impulse on stimulus type was observed ($p < 0.01$) (Fig. 5B).
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48 270 Note that, while impulse was precisely controlled among subjects, peak force
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50 271 exhibited some increased dispersion at 2 Ns compared to other impulse levels,
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52 272 possibly due to the difficulty in controlling short-duration perturbations.
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2 273 In all subjects, ΔCoP exhibited a significant ($p < 0.001$) and extremely good linear
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4 274 correlation with the impulse of the perturbation (Fig. 6A), $r = 0.96$ on average, in
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7 275 spite of the slight differences observed in average peak force levels. Conversely,
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9 276 the mean correlation between ΔCoP and force amplitude was poor ($r = 0.49$) and
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11 277 not statistically significant in 7 out of 14 subjects (Fig. 6B). The box plots of Fig.
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13 278 6C show the distribution of the individual Pearson's correlation coefficients in the
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17 279 two cases.

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20 280 The linearity of the relation between ΔCoP and impulse allowed normalizing the
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22 281 CoP displacement to the impulse of the perturbation: $\Delta\text{CoP}_n = \frac{\Delta\text{CoP}}{\text{Impulse}}$, which
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25 282 should then provide a postural index independent of the magnitude of perturbation
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27 283 (Dvir et al. 2020). This index remained fairly constant, within the constant-force
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29 284 series for impulse (range: 4-10 Ns). Friedman's ANOVA indicated a significant
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31 285 dependence of ΔCoP_n with impulse ($p < 0.01$) with a significantly increased value
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33 286 at impulse = 2 Ns compared to the other magnitudes ($p < 0.01$) (Fig. 7A). Also in
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35 287 the constant-impulse session, the experimental ΔCoP_n was influenced by the force
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37 288 amplitude of the perturbation ($p < 0.01$) but only the response to $F=100$ N differed
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39 289 significantly from the other magnitudes (Fig. 7B): the ΔCoP_n at 100 N was
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41 290 significantly higher than the ΔCoP_n at 20 N ($p < 0.05$) and at 40 N ($p < 0.01$).
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43 291 Notably, on exclusion of the low-impulse (2 Ns) and high-force perturbations (100
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45 292 N) the individual ΔCoP_n values remain fairly comparable, even in response to
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47 293 different stimulus types (ICC = 0.88 with 95% confident interval [0.75 – 0.96]).
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49 294 Furthermore, the normalized index ΔCoP_n showed relatively low variability when
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51 295 assessed in response to 5 perturbations of the same type: on average $\text{CoV} = 13 \pm 7\%$.
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2 296 A single index value was calculated for each subject by averaging the ΔCoP_n over
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4 297 all perturbations greater than 2 Ns and less than 100 N (mean [range]: 0.93 [0.72 –
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6 298 1.15] cm/Ns). The mean value of the ΔCoP_n was significantly inversely correlated
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8 299 with the physical characteristics of the subjects: weight ($r = -0.79$), height ($r = -$
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10 300 0.69) and foot length ($r = -0.63$).

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14 301 In order to exclude postural adjustments in preparation for back perturbations, the
15
16 302 resting CoP was analyzed for possible variations during the test. No significant
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18 303 change in resting CoP was detected within any of the 2 session and of the 8
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20 304 perturbation sequences.

21 22 23 24 25 26 305 *3.2 Simulations results*

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28 306 The tuning of the model was performed to match the average experimental ΔCoP_n
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30 307 response of Fig. 7A (black line). The comparison between simulation results and
31
32 308 experimental data, for each testing condition selected during the trials carried out
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34 309 on healthy subjects, is shown in Fig. 8.

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36 310 It can be observed that, in the absence of sensory noise, simulated ΔCoP exhibited
37
38 311 a linear trend with the impulse (Fig. 8A, blue line) whereas no dependence on the
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40 312 force amplitude (Fig. 8B) was found. Accordingly, ΔCoP_n remained extremely
41
42 313 constant over the entire range of impulse and force amplitude (Fig. 8C and D).

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44 314 With the addition of noise to the sensory feedback, both ΔCoP and ΔCoP_n increased
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46 315 in all conditions (Fig 8 A-D, red lines). While this effect was uniform for ΔCoP in
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48 316 all conditions, it was particularly marked at low impulse for ΔCoP_n , thus faithfully
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50 317 matching the experimental data at 2 Ns.

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2 318 **4. DISCUSSION**

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4 319 To the best of our knowledge, this is the first study in which force and impulse of
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6 320 the trunk perturbations have been systematically varied in order to investigate their
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9 321 differential effect on PR. The issue was addressed by challenging the balance of
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11 322 healthy subjects by means of a custom-built perturbator, which proved adequate to
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13 323 deliver accurately controlled stimuli, and by analyzing simulated responses based
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15 324 on a simple inverse pendulum model.

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19 325 The findings support the hypothesis formulated on the basis of a previous
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21 326 observation, namely, that the displacement of the CoP is consistently and strongly
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23 327 correlated with impulse and not significantly correlated with the force amplitude of
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25 328 the perturbation. Furthermore, since the extracted ΔCoP_n was quite constant across
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27 329 the perturbation range, the applicability of this index as a synthetic descriptor of the
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29 330 individual postural performance was further amplified.

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34 331 Although, as pointed out, the association between ΔCoP and the magnitude of the
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36 332 perturbation has been highlighted before, a clear *linear* relationship has been
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38 333 evidenced experimentally only in a handful of studies. Kim et al (2009) showed that
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40 334 ΔCoP was positively correlated with the peak force of perturbations applied to the
41
42 335 high back, in apparent contrast with the present results. However, we speculate that
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44 336 the duration of the perturbations (which was not measured) was quite constant
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46 337 across the different subjects, which would make impulse and force amplitude
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48 338 proportionally related and thus, both correlated with ΔCoP . Our preliminary study
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50 339 on PR (Dvir et al. 2020) indicated a moderate correlation between ΔCoP and force
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52 340 ($r = 0.50$) and a stronger correlation with the impulse of the perturbation ($r = 0.71$)

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2 341 but the distributions of the individual Pearson correlation coefficients were quite
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4 342 dispersed, possibly because the study was based on uncontrolled manually-
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6 343 delivered perturbations. The possibility to deliver accurate perturbations in the
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8 344 present study effectively reduced the intra-subject variability in the PR and revealed
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10 345 the clear-cut linear relationship between ΔCoP and impulse ($r = 0.96$) while
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12 346 confirming a low correlation between ΔCoP and force amplitude ($r = 0.49$ on
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14 347 average but reaching significance only in 7 subjects). Moreover, the reproducibility
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16 348 of the disturbances provided by the perturbator was adequate for the application, as
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18 349 signaled by the results shown in Fig. 5, confirming that the performance of the
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20 350 device was not significantly affected by the presence of a human operator (Ferraresi
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22 351 et al. 2020b; Maffiodo et al. 2020). Notably, as compared to our previous study
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24 352 based on manual uncontrolled perturbations, with the new perturbator we were able
25
26 353 to reduce the within-subject variability of ΔCoP_n , from about $20 \pm 8\%$ (recalculated
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28 354 from previous data) to $13 \pm 7\%$. As a result, it was here possible to achieve a
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30 355 comparable ICC with as few as 5 perturbations, instead of the 20 stimuli used in the
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32 356 previous study.

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42 357 The results of the study reinforce the concept that a single index, ΔCoP_n , obtained
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44 358 from the ratio of ΔCoP and impulse, may synthetically describe the PR of the
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46 359 subject, independently of the magnitude of the perturbation (Dvir et al. 2020). In
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48 360 fact, this index is here shown to remain fairly constant in a wide range of force and
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50 361 impulse intensity (Fig. 7). Notably, this index was slightly but significantly
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52 362 increased at low impulse and high force amplitude: a pattern not predicted by the
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54 363 model (Fig. 8 D). While significant non-linearities are embedded in the postural
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56 364 control system, starting from the muscle level (Ivanenko and Gurfinkel 2018), the

1
2 365 present deviation from linearity could be related to the short duration of the
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4 366 perturbation, which is below 75 ms for both 2 Ns and 100 N. In fact it has been
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6
7 367 proposed that short stimuli elicit a triggered response, uninfluenced by the stimulus
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9 368 characteristics, while a longer stimulus duration would be necessary for sensory
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11 369 inputs to encode the magnitude of the perturbation and help to shape a proportionate
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13 370 response (Diener et al. 1988). On the other hand, the results here obtained with the
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15 371 model also suggest that, at low perturbation magnitudes, the presence of noise in
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17 372 the system may account for a similar non-linearity (Fig 8 C-D).
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23 373 While the implemented model completely excludes a dependence of the PR on the
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25 374 force amplitude, a significant correlation was evidenced in some subjects (Fig. 6B).
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27 375 It may be observed that these individual correlations are based on only 4 points and
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29 376 thus heavily depend on each single measurement. As a consequence, increased
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31 377 correlations would result due to the abnormally increased response at 100 N, as
32
33 378 previously discussed. On the other hand, a weak correlation with the force
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35 379 amplitude could also result from the involvement of additional sensory feedback
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37 380 pathways, particularly sensitive to the force stimulus (e.g., touch receptors of the
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39 381 back, vestibular receptors), not included in the present model.
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46 382 Regarding the accuracy of the simulations, the approach to model tuning used in
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48 383 this study was considered suitable to achieve a realistic although simplified
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50 384 behavior of the model, however it is well known that all the active and passive
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52 385 response parameters discussed are highly subject-specific and require accurate
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54 386 estimation when a detailed description of balance control is targeted (Goodworth
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56 387 and Peterka 2018).
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2 **388 5. LIMITATIONS**
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5 389 As a first approximation, the balance reaction of healthy young adults in response
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7 390 to low disturbance mainly consists of a correcting torque at the ankle (Horak and
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9 391 Nashner 1986; Shumway-Cook and Woollacott 2007). Therefore, a single-link
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11 392 inverted pendulum model was developed to simulate the postural response of the
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13 393 study participants. This approximation was supported by the visual inspection of
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15 394 the experimental trials, that confirmed how most oscillations occurred about the
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17 395 ankle joints. As indicated by the good match between experimental and simulated
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19 396 data, this simple model proved to be sufficiently accurate for the purpose of testing
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21 397 the relationship between the displacement of the CoP and the impulse of the
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23 398 perturbation. On the other hand, we cannot exclude that other postural strategies,
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25 399 such as the hip strategy, could also contribute to the whole response, particularly to
26
27 400 high-magnitude perturbations. This would likely affect the correlation between
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29 401 ΔCoP_n and impulse, although the precise effects are difficult to predict, based on
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31 402 the present experiments. Appropriate integration of the hip strategy into the model
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33 403 requires to adopt a double-link inverted pendulum model, resulting in a far more
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35 404 complex optimization problem, with additional unknown control parameters used
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37 405 to model the correcting torque at the hip and the interaction between active controls
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39 406 at each joint (Goodworth and Peterka 2018). This, in turn, requires the acquisition
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41 407 of additional descriptors of the postural response, e.g. tangential forces at the
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43 408 platform, movements and acceleration of the different body segments. The present
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45 409 results suggest that this increase in complexity is not necessary for describing the
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47 410 response to small postural perturbation.
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2 411 Another limitation of the study was the non-exactly constant value of the force
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4 412 amplitude and of the impulse in the force constant session and in the impulse
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6 413 constant session, respectively. The perturbations were applied to the subjects with
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8 414 a custom-made device consisting of a low friction pneumatic actuator controlled in
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10 415 force and position by a PI controller. The nonlinearities and relatively slow
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12 416 dynamics associated to pneumatic systems and the inertia of the piston make the PI
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14 417 controller not able to appropriately minimize the error between the force reference
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16 418 profile and the applied force in a very short time. As a result, there is an overshoot
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18 419 in the first 35 ms of the perturbation that impacts on the calculated Force Amplitude,
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20 420 especially in the case of short-lasting perturbations. To obtain more accurate
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22 421 perturbation profiles and more robust control, an electrically-actuated perturbator
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24 422 based on Model Predictive Control, with inherent high dynamics and stiffness, is
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26 423 currently under development (Pacheco Quiñones et al. 2021).
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36 424 **6. CONCLUSION**

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39 425 The results support the use of the impulse rather than the force as input variable in
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41 426 impulsive perturbations applied to the trunk. Thanks to the linearity of the
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43 427 relationship between ΔCoP and impulse, the postural index, ΔCoP_n , may be used
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45 428 as a synthetic descriptor of the individual postural performance.
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50 51 429 **CONFLICT OF INTEREST STATEMENT**

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54 430 The authors have no conflict to disclose.
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58 59 431 **ACKNOWLEDGMENTS**

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435 the study.

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31 **FIGURE LEGENDS**
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34 552 **Figure 1.** Experimental Set-up. A: pneumo-tronic perturbator, 1: low friction
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36 553 pneumatic actuator, 2: flow-proportional valves, 3: laser sensor, 4: load cell,
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38 554 5: end striker, 6: handles, 7: trigger button. B: Example of experimental task
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40 555 with the operator handling the pneumo-tronic perturbator.
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45 556 **Figure 2.** Force profiles for the different perturbation types included in the constant
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47 557 force series (A) and the constant impulse series (B). The intended force
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49 558 profile (red) is superimposed to the actually delivered force profile (blue,
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51 559 average across all subjects).
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55 560 **Figure 3.** A representative recording of the perturbation (Black line) and the
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57 561 ensuing displacement of the Center of Pressure (dashed grey line) observed
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2 562 during experimentation (constant-force series: 40 N, 6 Ns).
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5 563 **Figure 4.** Free body diagram of a single-link inverted pendulum model for postural
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7 564 control analysis. θ is the body oscillation, l is the height of the subject with
8 565 respect to the ankle joint; h_F is the distance between ankle joint and the point
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10 566 of application of the perturbation force F_e ; d is the distance between ankle
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12 567 joint and the center of mass (CoM); h is the height of ankle joint with respect
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14 568 to the fixed base of support; I is the rotational inertia of the body about the
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16 569 CoM; m is the subject body mass; \ddot{x} is the horizontal acceleration of the CoM;
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18 570 \ddot{y} is the vertical acceleration of the CoM; $\ddot{\theta}$ is the angular acceleration of the
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20 571 CoM; g is the gravitational acceleration; τ is the correcting torque at the
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22 572 ankle; CoP is the center of pressure position; R_x is the horizontal component
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24 573 of the ground reaction force; R_y is the vertical component of the ground
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26 574 reaction force
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35 575 **Figure 5.** Characteristics of delivered perturbations for the constant-force series
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37 576 (left) and the constant-impulse series (right). Each box represents the median
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39 577 and the standard deviation of the perturbations applied to the subjects (n=5
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41 578 perturbation x 14 subjects = 70), for each stimulus type.
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46 579 **Figure 6.** The relationship between the maximum displacement of the center of foot
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48 580 pressure, ΔCoP , and the magnitude of the perturbations, in terms of impulse
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50 581 (A) and force amplitude (B) for each participant in the experimental trial.
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52 582 Distribution of the Pearson's Correlation Coefficients, for the ΔCoP –
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54 583 Impulse (Black) and the ΔCoP - Force (white) correlation (C).
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2 584 **Figure 7.** The relationship between the postural index ΔCoP_n and the magnitude of
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5 585 perturbation expressed in terms of impulse (A) and force amplitude (B) for
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7 586 each participant in the experimental trial (colored line). The thick black line
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9 587 represents the average trend.

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12 588 **Figure 8** The relationship between the simulated maximum displacement of the
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15 589 center of foot pressure, ΔCoP , and the magnitude of the perturbations, in
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17 590 terms of impulse (A) and force amplitude (B). The relationship between the
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19 591 postural index ΔCoP_n and the magnitude of perturbation expressed in terms
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21 592 of impulse (C) and force amplitude (D).
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23 593 Red lines refer to the results of the simulation performed considering the
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25 594 sensorial noise; blue lines refer to the results of the simulation performed
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27 595 without the contribution of the sensorial noise; black lines are the average
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29 596 experimental trend calculated on all the participants of the experimental
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31 597 analyses.
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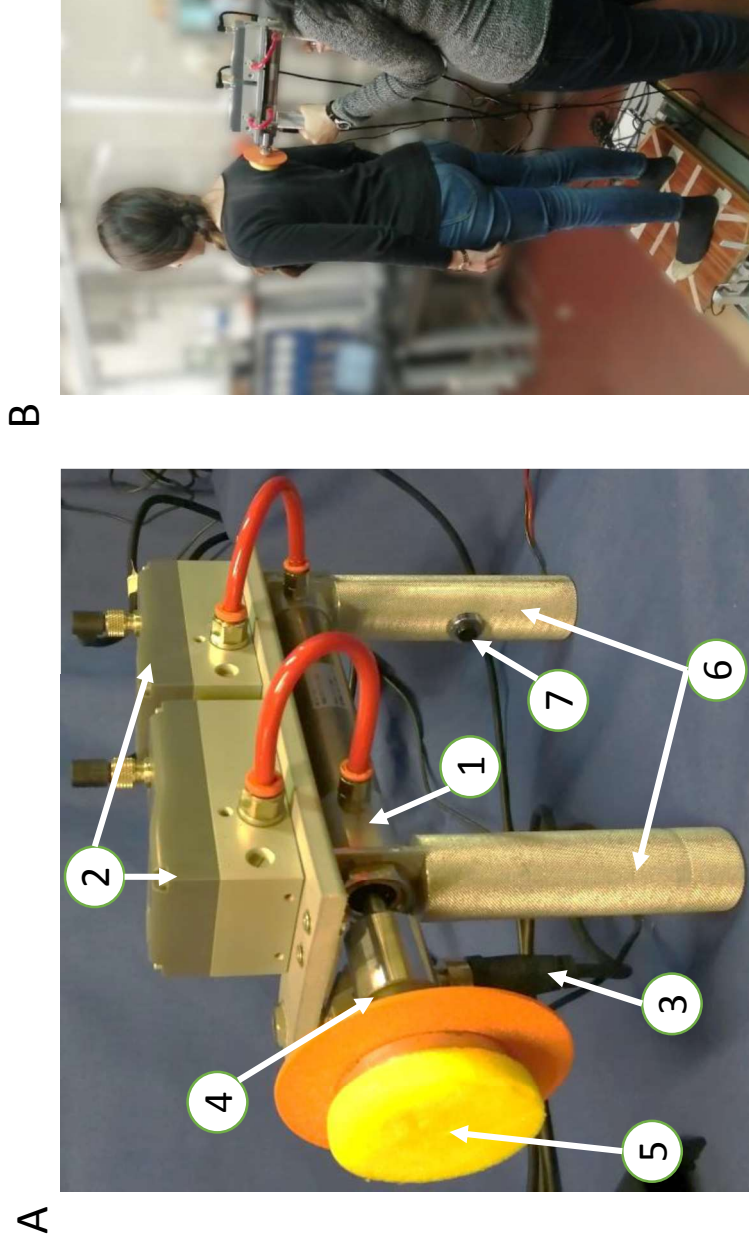


Figure 1

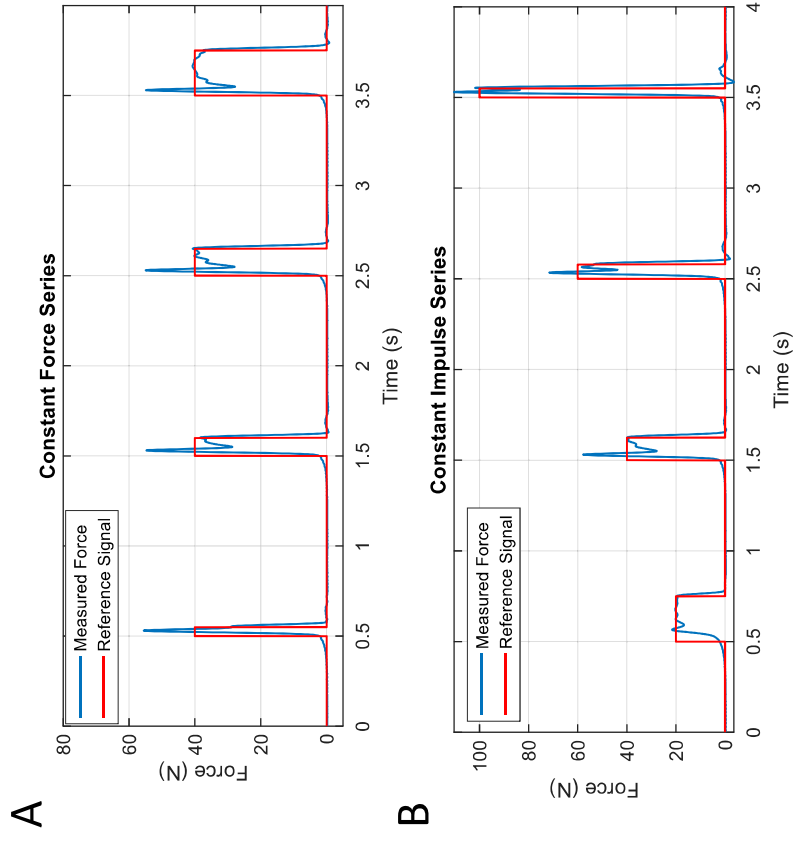


Figure 2

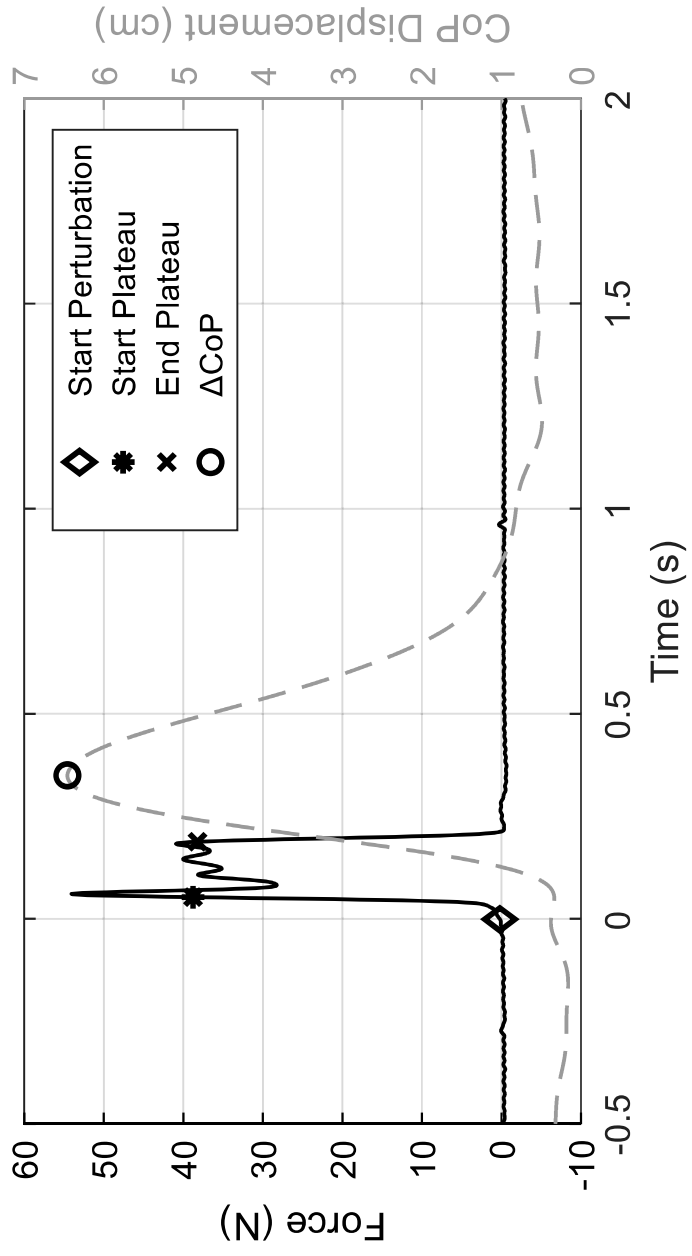


Figure 3

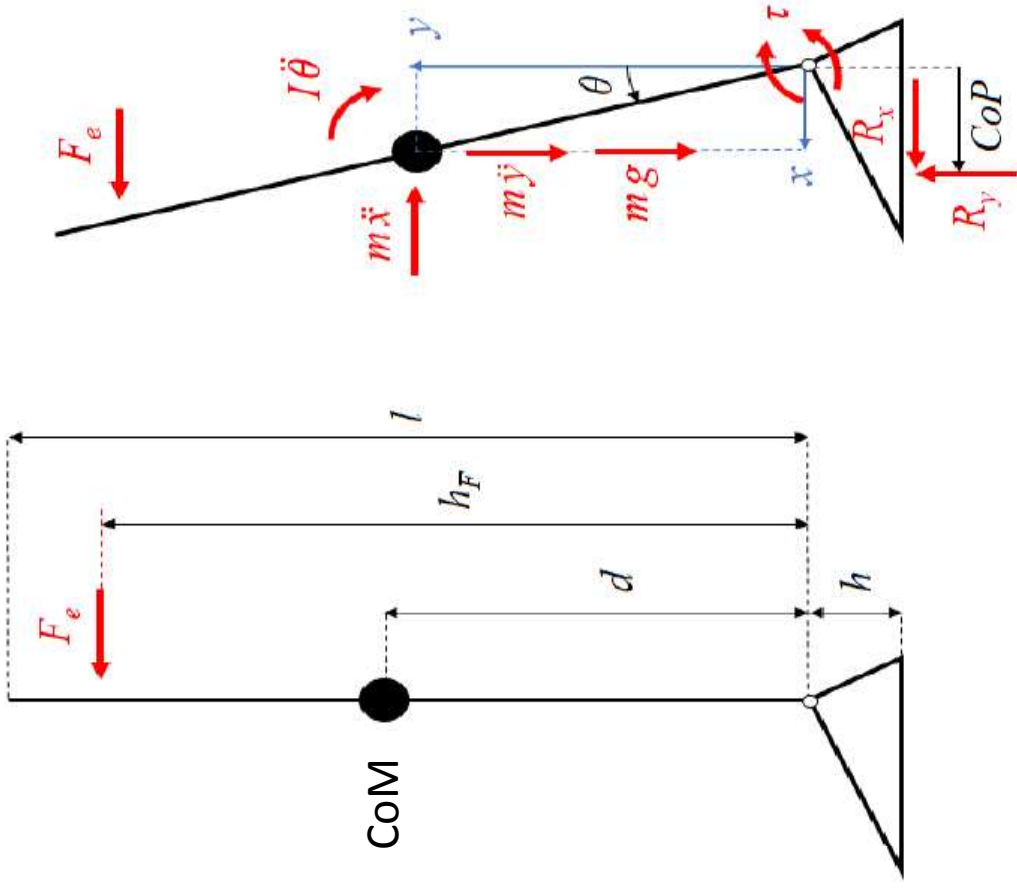


Figure 4

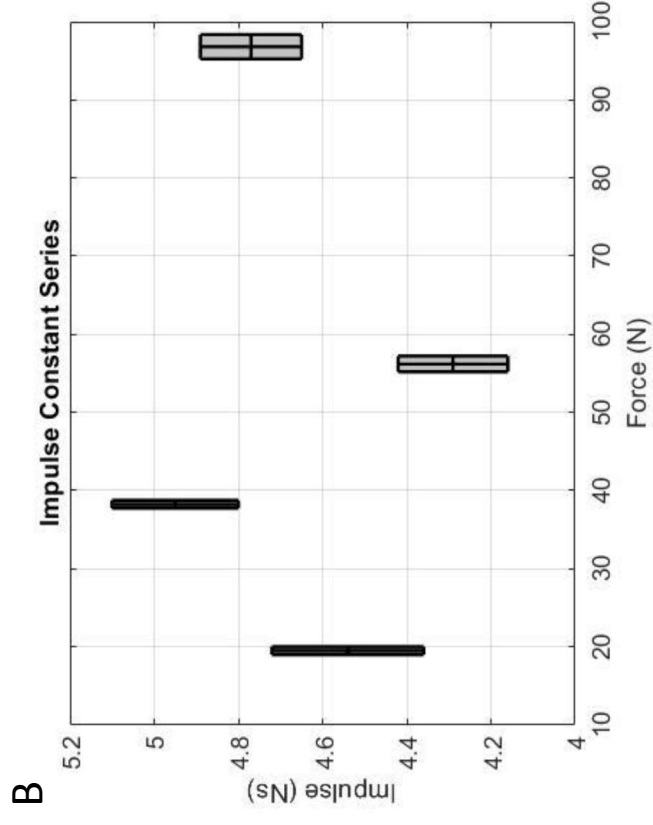
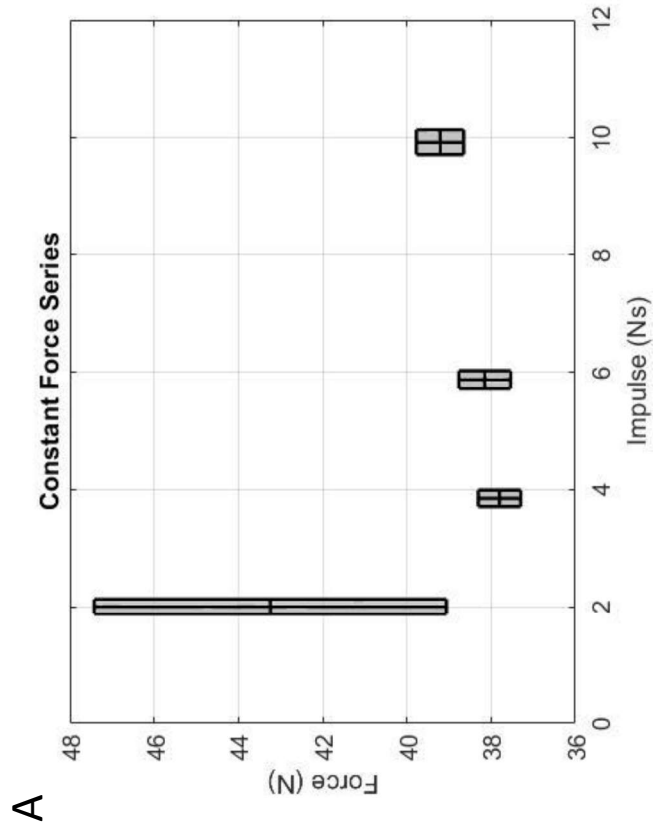


Figure 5

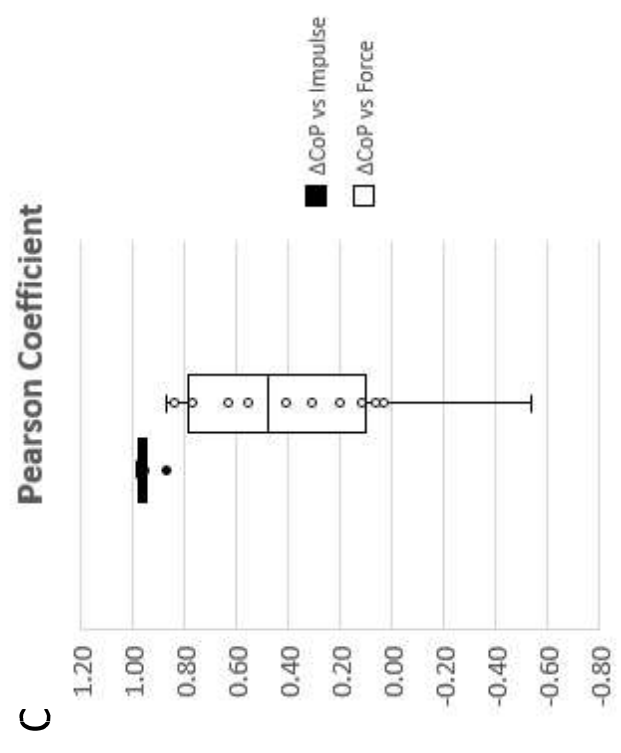
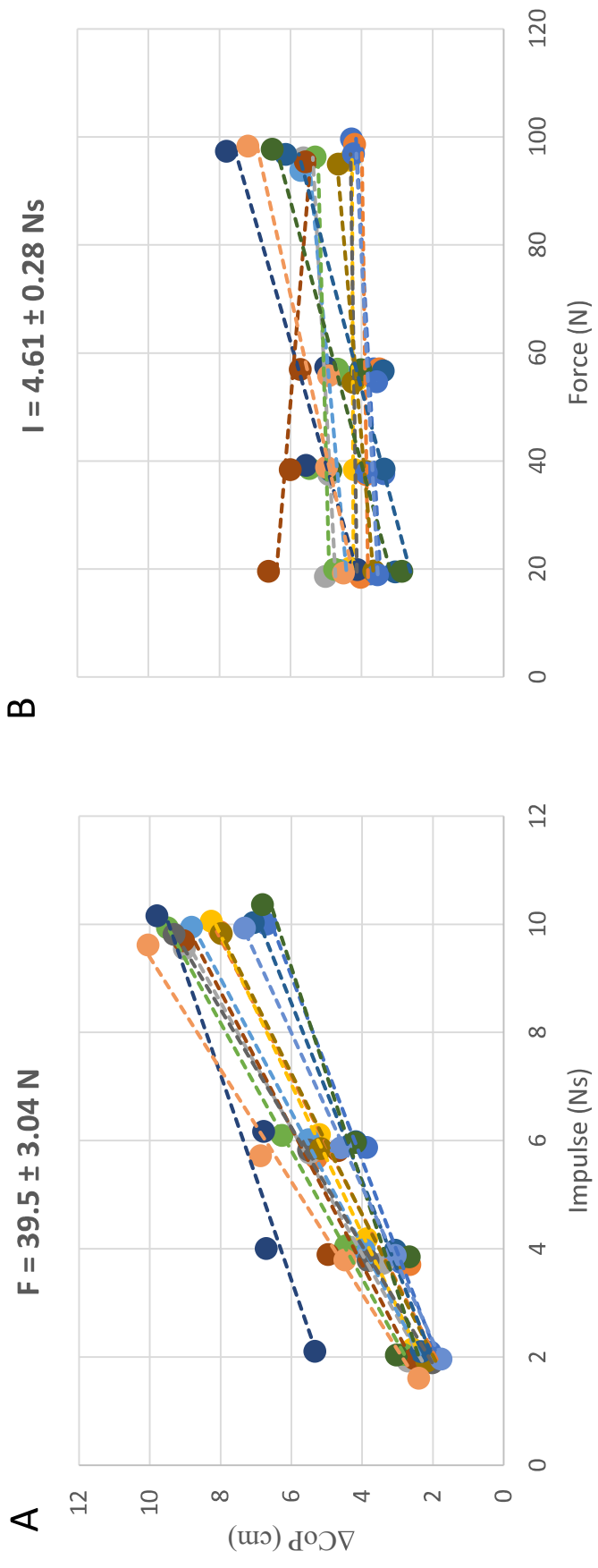


Figure 6

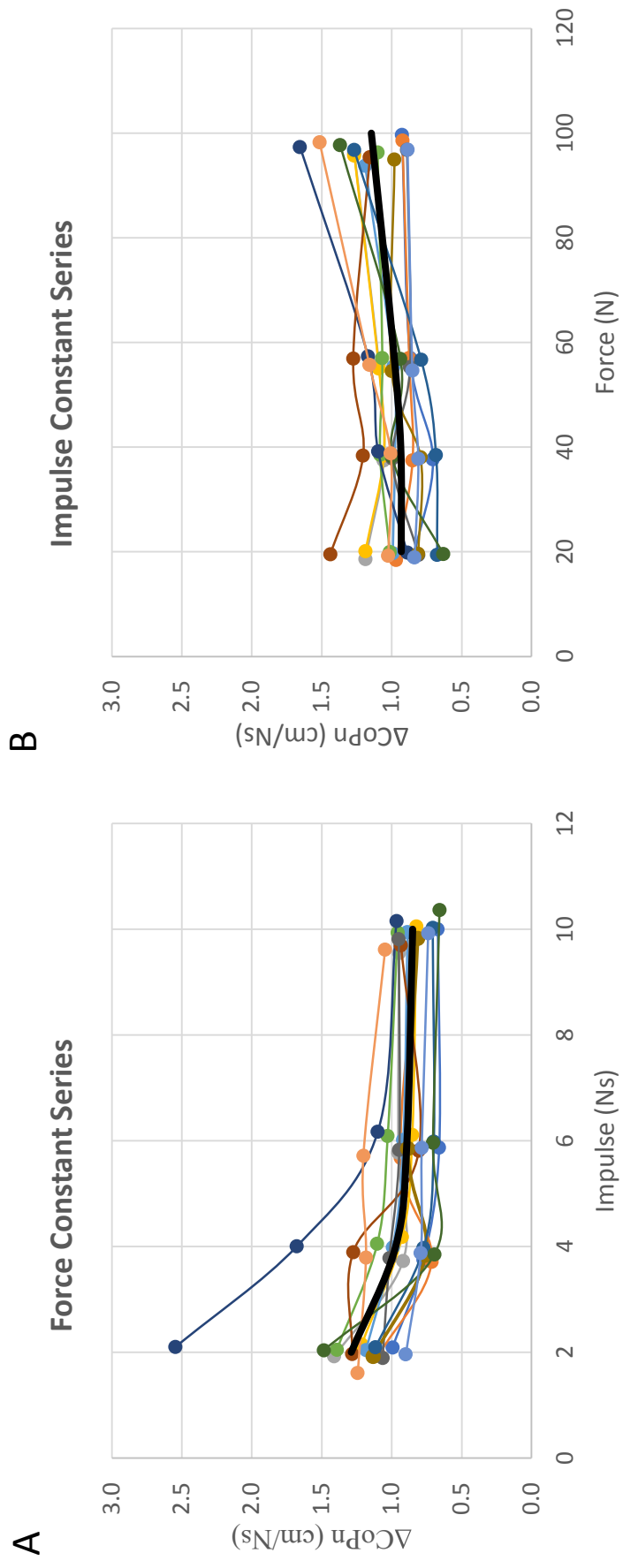


Figure 7

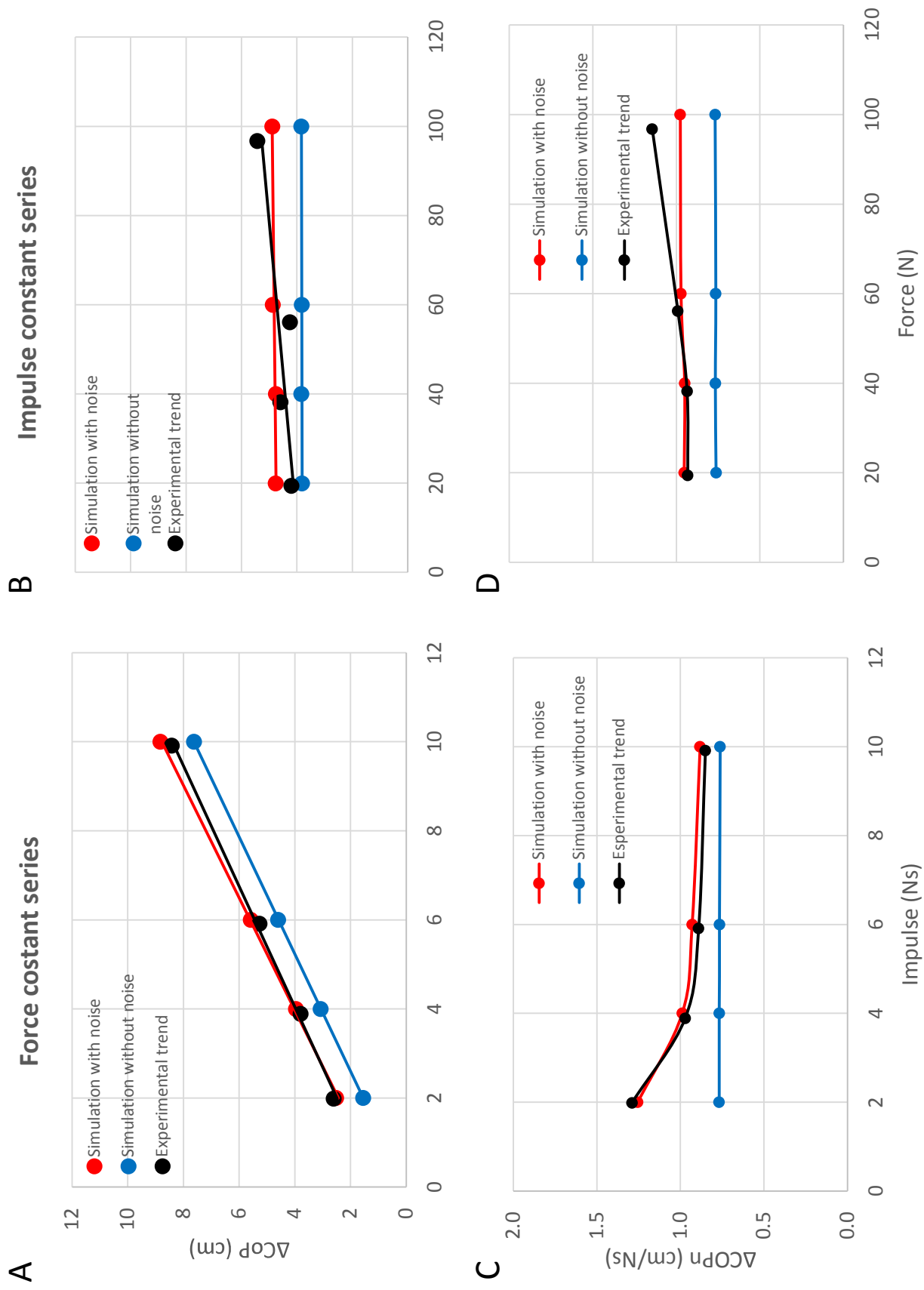


Figure 8