Linearity and repeatability of postural responses in relation to peak force and impulse of manually delivered perturbations: a preliminary study

Linearity and repeatability of postural responses in relation to peak force and impulse of manually-delivered perturbations: A preliminary study

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All authors agree to be accountable for all aspects of the work in ensuring that questions related to the accuracy or integrity of any part of the work are appropriately investigated and resolved.

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ABSTRACT

PURPOSE Postural reactions of standing subjects have been mostly investigated in response to platform displacements or body perturbations of fixed magnitude. The aim of this study was to investigate the postural response as a function of peak force and impulse of the perturbation.

METHODS In 10 healthy young men, standing balance was challenged by anteriorly-directed perturbations (peak force: 20-60 N) delivered to the back, at the lumbar (L) or inter-scapular (IS) level, by means of a manual perturbator equipped with a force sensor. Postural responses, in terms of displacement of the center of pressure (CoP), were recorded by a force platform. Two sets of 20 randomly ordered perturbations (10 to each site) were delivered in two separate testing sessions. RESULTS The magnitude of CoP response (ΔCoP) was better correlated with the impulse (I) than with the peak force of the perturbation. The normalized response, $\Delta CoP_n = \Delta CoP / I$, exhibited good reliability (ICCs of 0.93 for IS and 0.82 for L), was higher with IS than with L perturbations ($p < 0.01$) and was significantly correlated with the latency of CoP response: $r = 0.69$ and 0.71 for IS and L, respectively.

CONCLUSION These preliminary findings support the concept that manually delivered perturbations can be used to reliably assess individual characteristics of postural control and that ΔCoP_n may effectively express a relevant aspect of postural control.

Keywords: posture; balance perturbation; center of pressure; postural reflex.

List of abbreviations

ANOVA	Analysis of variance
CoP	Centre of pressure
I	Impulse
ICC	Intra-class correlation coefficient
IS	Inter-scapular
L	Lumbar
PF	Peak force
PR	Postural reaction
SD	Standard deviation
t_0	Time at which the perturbation starts
t_e	Time at which the perturbation ends

INTRODUCTION

Postural reactions (PR) to mechanical perturbations have been the subject of a growing body of research in recent years. Unless meant to test anticipatory reactions, these perturbations are imparted unexpectedly, threatening to move the body out of its equilibrium. Since the pioneering study by Horak and Nashner (1986) a large number of studies adopted the approach of perturbing the base of support, which simulates situations like a slip motion or the acceleration of a bus. Noteworthy, postural challenges may often arise from unexpected perturbations directed to the body, as may occur during collisions with other people or objects. However, a relatively smaller number of studies used an experimental model to study such perturbations which may in fact elicit different PR patterns compared to those obtained using sudden movement of the base of support (Colebatch et al. 2016; Chen et al. 2017). This is likely related to the difficulty in providing a standardized body perturbation, which by its very nature is a vector and hence is characterized by its location, magnitude and direction, developing over a short time interval.

To study this kind of PR due to body-directed perturbations different approaches have been proposed. In some cases, pretty complex systems have been implemented allowing to impart perturbations from different directions by the aid of suspended weights or electric actuators. For instance, Sturnieks et al (2013) investigated step reactions in elderly subjects by implementing motor-generated perturbations, imparted to a standing subject through 4 cables tethered to a waist harness and extending anteriorly, posteriorly, to the left and to the right (coronal plane). Forghani et al (2017) implemented a robotic joystick driven by electrical actuators to investigate postural reactions on trunk and limb muscles to multi-directional perturbation imparted to the subject’s arm.

Besides the use of linear motors (Pidcoe and Rogers 1998; Gilles et al. 1999; Mille et al. 2003) postural perturbations have also been delivered by suddenly applying loads, pulling the body through flexible cables connected to harnesses or rigid vests (Cresswell et al. 1994; Martinelli et al. 2015; Di Giulio et al. 2016) or to handles or balloons held by the subjects (Aruin and Latash 1995; Piscitelli et al. 2017). In some instances the perturbation could result from suddenly releasing a load that was previously resisted by the subject (Aruin and Latash 1995; Piscitelli et al. 2017). A disadvantage of these systems is the fact that the subject is somewhat constrained by the harness and cables or is actively engaged in resisting a load or in holding a handle.

Other approaches left more freedom to the subjects. One experimental model was based on releasing a pendulum which hits the body at shoulder level (Santos and Aruin 2009; Chen et al. 2016). The pendulum's length could be adjusted according to the subjective shoulders' height while its weight could be set according to the desired magnitude of the impact, e.g. equal to a certain percentage of the subject's body weight (Chen et al. 2016). An even simpler model has been recently proposed in which the perturbation was manually imparted by the experimenter who had placed his hand on the subject's shoulder and briefly pulled the latter's torso in the anterior or posterior direction (Colebatch et al. 2016; Colebatch and Govender 2019). One important limitation of this experimental design is that the mechanical perturbation is poorly characterized in terms of onset, magnitude and time course. In some case the ensuing body acceleration has been used as a surrogate (Colebatch et al. 2016; Chen et al. 2017; Colebatch and Govender 2019), although it was actually the output variable of the perturbed system.

A variant of this approach is described in the present study which overcomes some of the limitations outlined above: a simple device which acts as an extension of the experimenter's hand is used to provide moderate perturbations in the form of a push force to the back of subjects who stand on a force platform. A similar approach has been previously adopted to investigate joints kinematics (Kim et al. 2009) or balance control in patients with Parkinson's Disease (Pasman et al. 2019) but without addressing the relation between magnitude of perturbation and of the postural response.

Notably, irrespective of the adopted approach, the magnitude of the perturbation is generally characterized in terms of force, although it is the impulse of the perturbation (= integral of force over time) that quantifies the momentum transferred to the body. Moreover, the relation between the magnitude of the perturbation and resulting displacement of the center of pressure (CoP) appears to have been addressed only in a handful of studies relating especially to the threshold for step reactions (Mille et al. 2003; Robert et al. 2018; Le Mouel et al. 2019).

The principal aim of the present study was therefore to explore the relationship between anteriorly-directed perturbations to two distinct markers on the back and the specific PR expressed by displacement of the CoP. We hypothesized that this displacement would be roughly proportional to the magnitude of the perturbation and better correlated with the impulse than with the peak force. Furthermore, testing the subjects at two distinct sessions was intended to yield a preliminary idea about the repeatability of this specific PR, an issue which has not been investigated in this context, hitherto.

METHODS

Subjects

A group of 10 healthy young adult men (mean[range] age: 27.3 ± 6.1 year [25-35y]; height: 1.76 ± 0.07 m [1.65-1.88 m]; weight: 69 ± 9.38 kg [60 – 86 kg]) was recruited using a notice board. None presented with orthopedic, neurological or any other general physical problems. All subjects provided written informed consent to participation in this research, which has been approved by the institutional review board of the University of Torino (#360583, November 2017). All procedures were conducted in accordance with the Helsinki declaration.

Instrumentation

The testing set-up comprised two instruments: a manual perturbator and a force platform.

The perturbator (Fig. 1) was designed and constructed at the Dept. of Mechanical and Aerospace Engineering at the Politecnico di Torino and consisted of the following elements:

1. A handle, made with a 3D printer for maneuvering the device;
2. A uniaxial load cell (Dacell UMM, Korea, rated capacity about 500 N);
3. A load cell-subject interface element lined with deformable synthetic foam (diameter = 6.6 cm; thickness = 1 cm; density = 22.3 Kg/m^3)

The force platform, a modified Shekel (Beit Keshet, Israel) device, consisted of an upper plate (52x36 cm) which was supported by 4 uniaxial load cells (TEDEA, Israel, model 1042, rated capacity 100 kgf). From the distribution of the forces, the antero-posterior and medio-lateral position of the CoP could be calculated.

A conceptual model of the experimental set-up is shown in Fig. 2. The force signals from the perturbator and from the load cells of the force platform were conditioned and acquired by 16-bit A/D converter (Micro1401-mkI, CED, UK) at 1000 Hz, with Spike II acquisition software (CED, UK).

Procedure

Testing took place at the Lab of Integrative Physiology, Dept. of Neuroscience, the University of Torino. Prior to the tests, the operator who performed all criterion tests, was trained in delivering horizontally directed perturbations within the range of 20-60 N. Based on preliminary experiments, this range was set in order to obtain clear postural responses while excluding the risk of eliciting stepping responses.

Given the design of the apparatus the perturbations could not be weight- or height-normalized. The exact location of the perturbation was ensured by sticking round adhesives on the hitting point. During the test, the subjects stood comfortably barefoot on the force platform with the feet at pelvic distance and with vision unobstructed (Fig. 2). Subjects were asked to assume normal relaxed stance. The operator stood behind the subject holding the perturbator in their dominant hand while the interface was maintained at a distance of about 5 cm from the subject's back. This short distance, the horizontal direction of the perturbator and the alignment of the perturbation sites meant that the force recorded by the perturbator was equal or very nearly so to the actual force imparted to the subject's back. Subjects were first familiarized with the procedure experiencing a few perturbations while standing on the force platform. The subjects were told that the perturbations posed no risk of falling and were asked to behave naturally when they were imparted. Then, each subject received 40 perturbations to the trunk divided in two series of 20, with a break of 5 min in between. An inter-perturbation pause of at least 10 s was allowed for returning to relaxed stance. In each series, 10 perturbations were imparted at the inter-scapular (IS) level ($P_{I\text{-}S}$, between the vertebral angles of the scapulae) and 10 at lumbar (L) level (P_L, at the level of L3), at an intensity roughly spanning the intended 20-60 N force range. Having two points of application of the perturbation to the back allowed us to limit anticipatory reactions by the subject, who knew neither the point of application nor the exact instant of the perturbation, without altering the sensorial inputs of the subject, for example with shielded glasses. No further points were considered in order limit the overall duration of the experimental protocol. A random sequence of IS and L perturbations was initially generated and then used in all session and all subjects. The perturbations were imparted by the examiner without the subject being able to anticipate the site or the magnitude. A typical testing session lasted 15 min. The same sequence of perturbations was repeated in a second session, 3-6 days later.

Measured variables

The main variables explored in this study are depicted graphically in Fig. 3. The force signal of the perturbation (Fig. 3, top) was filtered using a Butterworth low pass filter (4^{th} order, cut-off frequency 150 Hz), designed in MATLAB_R2016b®. The perturbation was characterized in terms of:

- Peak Force (PF, in N), the maximum value of the force recorded during the perturbation;
- Impulse (I, in N•s), the integral of force over the time interval delimited by the start (t_0) and the end (t_e) of the perturbation. For each perturbation t_0 and t_e were automatically detected as the time instants at which the force signal crossed a threshold equal to 5% of the PF, before and after the peak, respectively.
- Duration (in s) of the perturbation was computed as t_e-t_0.

CoP displacement in response to the perturbation (Fig. 3, bottom) was calculated according to the following formula:

$$CoP_{dis} = \sqrt{(AP - \overline{AP})^2 + (ML - \overline{ML})^2}$$

based on the low-pass filtered (Butterworth 4th order, cut-off frequency: 20 Hz) antero-posterior (AP) and medio-lateral (ML) coordinates of the CoP, $\overline{AP}$ and $\overline{ML}$ being calculated over a time interval of 3 s preceding the perturbation.

The postural response was characterized in terms of:

- Latency (in ms), was defined as the time interval from the start of perturbation to the start of CoP response which corresponded to the point in time at which the CoP displacement exceeded a threshold equal to its mean value + 2SD,
- ΔCoP (in cm) was calculated as the maximal value of the CoP displacement during the first second after the perturbation, as corroborated by visually inspecting the trace of the CoP and ensuring that all maxima were captured within this time frame.
- Time to Peak (in ms), the time interval from the start of CoP response to the instant at which the CoP reached the maximum displacement
- Duration of CoP response (in s), the time interval from the start of CoP response to the first local minimum of CoP after the peak.

Statistical Analysis

All statistical procedures were conducted using MATLAB_R2016b®. (MathWorks, USA). A two-way repeated-measures analysis of variance (ANOVA) with a grouping factors for site and or for day was performed on each of the variables characterizing both the perturbation and the response. For assessing the correlation between variables characterizing single perturbations and responses, within individual subjects, Pearson's r was applied. The Fisher's Z transform was then used to estimate an average correlation coefficient over all subjects, while the Wilcoxon non-parametric test was performed to compare correlations between different pairs of variables and to evaluate whether the response of the subjects varied significantly during the test. Data in the text are expressed as mean ± standard deviation, unless otherwise specified.

The intra-class correlation coefficient ($ICC_{3,k}$), based on a mean rating (k=20), absolute agreement, 2-ways mixed effects model (Koo and Li 2016), was used to quantify the reliability of the CoP response, in the two sessions. In addition, a mixed model analysis was applied, which integrates the data points of all subjects into a single model. A random intercept and slope model was applied to study the relationship between ΔCoP and the perturbation. The fixed effects were *I*, PF, Site (IS vs. L), Day (day1 vs day2 of the tests), the interaction between Site and *I* and the interaction between Site and PF. We have considered a random intercept and slope model for the different subjects.

However, prior to applying the mixed model analysis, the ΔCoP was log-transformed to obtain a normal distribution.

RESULTS

Characteristics of the perturbation

Only 7% of the perturbations fell outside the intended range of 20-60 N but were not excluded from the analysis. The perturbations did not depend on site and day, in terms of PF, *I* and duration. After averaging over all factors, the mean [range] values for these parameters were PF: 42.0 N [19-88 N], *I*: 2.10 Ns [0.62-5.54 Ns] and duration: 139 ms [45-353 ms] but only 8% of the perturbations lasted longer than 200 ms. All perturbations were very well tolerated by the subjects (verified by their verbal response) and no stepping reaction was ever observed.

Characteristics of the response

A representative tracing of the CoP response to the perturbation in the time domain is shown in Fig. 3. Noteworthy, the CoP response was delayed with respect to the start of the perturbation (t_0). The response latency was not dependent on the day but was significantly dependent on the site of stimulation resulting in 147 ± 56 ms and 97 ± 22 ms for the IS and L perturbations, respectively ($p<0.01$). On average the Time to Peak was 516 ± 250 ms and the whole response lasted 1.66 ± 0.36 s, both values being independent of site and day. Moreover, on average, the ΔCoP was not dependent on day but was significantly affected by perturbation site, 3.46 ± 1.42 cm and 2.53 ± 1.78 cm for the IS and L sites, respectively ($p<0.05$).

Derived relationships

The magnitude of the postural response is obviously dependent on the magnitude of the perturbation. However, to answer the question which parameters of the perturbation, the PF or the *I*, is more closely correlated with ΔCoP, the findings from a representative subject are first shown by scatter plots in Fig. 4 for the IS and L sites. Clearly, the ΔCoP was better correlated with the impulse (Fig. 4, right) than with PF (Fig. 4, left). The better correlation with the impulse was confirmed in 10/10 subjects for interscapular perturbation ($p<0.01$) and in 8/10 subjects for lumbar perturbations ($p=0.08$). Furthermore, the distribution of the Pearson correlation coefficients in the different conditions is shown in Fig. 5. On average, the correlation between ΔCoP and the PF was 0.50 ($r^2 = 0.27$) for both the IS and L perturbations but increased to 0.71 ($r^2 = 0.52$) and 0.67 ($r^2 =0.46$) when correlated with the *I*, respectively.

Based on the proportional relationship between ΔCoP and I, a new ratio: the normalized ΔCoP (ΔCoP_n) has been introduced:

$$\Delta CoP_n = \frac{\Delta CoP}{I}$$

In order to check whether there was any anticipatory activity in the postural responses we tested if the resting CoP and the ΔCoP_n changed during the series. We have observed that on average the resting position of the CoP measured in the last 30 seconds moved back by 0.63 ± 1.02 cm with respect to the first 30 s of the series. However, this difference was significant only in one of the four series (Day 2 – Series 1; $p<0.01$). Moreover, no significant changes in the ΔCoP_n were observed between the first and the last PR in each series. The bar diagram in Fig. 6a and in Fig.6b shows the mean value of ΔCoP_n over all perturbations, for the different subjects and conditions. This index exhibited a considerable variability across subjects and was systematically higher with IS (mean [range]: 1.68 [1.17 –2.97] cm/Ns) than with L (1.20 [0.74 - 2.35] cm/Ns) perturbations ($p<0.01$), although the ranking of the different subjects remained approximately the same in the two conditions with a Pearson's r of 0.94 ($p<0.001$, N=10, Fig. 6c). In terms of other relationships, the ΔCoP_n was negligibly correlated with the time to peak, with r = 0.15 and -0.02 for IS and L perturbations, respectively. Likewise, the r values for the correlation between ΔCoP_n and duration of the PR were 0.49 (IS, $p<0.05$) and 0.36 (L, p=0.12). On the other hand, ΔCoP_n was moderately but significantly correlated with the latency of the PR: r = 0.69 (IS, $p<0.05$) and 0.71 (L, $p<0.05$) (Fig. 6d).

Reliability

Using intra-class correlation coefficients ($ICC_{3,20}$) for determination of the test-retest reliability, the values for ΔCoP_n were 0.93 and 0.82 for IS and L perturbations, respectively. The latency of the postural response was also equally reliable: ICC = 0.94 and 0.90, respectively.

Mixed Model Analysis

Using this statistical method yielded a prediction formula ($p<0.05$) associating the output and input parameters as follows: $\ln(\Delta CoP) = 0.051$ (if Day2) – 0.318 (if Lumbar) $+0.276*I + 0.007*PF$

Transformation of this logarithmic formula indicated that the relationship between the perturbation and the displacement of the CoP was nearly linear.

DISCUSSION

Recognizing both the singular importance of understanding human PR to mechanical perturbation and the need to standardize the relevant tests in conjunction with the use of affordable, simple yet quantifiable instruments have motivated this preliminary study. For realizing the latter objective, we have designed and constructed a manual perturbator that enabled a precise measurement of the magnitude of the perturbation vector which was synchronously acquired with the Centre of Pressure, measured by the force plate.

The main findings of this preliminary study are as follows: 1) the postural response as expressed by the ΔCoP (the anterior maximal displacements of the CoP) is proportional to the magnitude of the perturbation and hence allows treatment of the body as a linear system, within the tested range; 2) this input-output relation is even more linear if the magnitude of the perturbation (the input) is expressed in terms of impulse rather than of peak force; 3) The resulting postural index $\Delta CoP_n = \Delta CoP/I$, exhibited high repeatability across different days and was correlated with the latency of the postural response; 4) Perturbations to the IS region produced similar responses compared with the L region but larger in magnitude and latency. These results, which were further corroborated by a mixed model analysis, support the applicability of this assessment of postural control while begging for some elaboration.

Another aspect of this study concerns the possible role of anticipatory postural adjustments (APA) in modifying PR, an issue that has been explored in numerous studies (Latash and Hadders-Algra 2008). Although some general mechanisms have been elucidated, the diversity of testing protocols (Bortolami et al. 2003; Le Mouel et al. 2019) as well as the nature of the perturbation (Piscitelli et al. 2017) generally do not allow direct comparisons between findings. However, in order to assess the results, it was necessary to find out whether APAs have actually been present during the test and if they did, to explore their effect. Notably, although the perturbations were imparted randomly (to I-S and L sites), with a reasonably long inter-perturbation pause and without any preceding auditory cue, the presence of APAs could not be precluded. However, by comparing the beginning and the end of each experimental series we verify that resting CoP was only slightly modified while ΔCoPn remained unaffected, suggesting that no discernible APA took place during the test protocol.

The manual perturbator and the test protocol

Due to its low weight and comfortable grip, the manual perturbator could be easily manipulated by the operator; its large impact surface produced no discomfort to the subjects. By design, it was little sensitive to non-axial loads and provided good quality signals for accurate description of the perturbation. Description of similar device was briefly reported (Kim et al. 2009; Pasman et al. 2019) but not further developed. Other tested approaches to deliver external perturbations to the trunk related to either complex instruments (Mille et al. 2003; Mansfield and Maki 2009; Ayena et al. 2016) or did not provide precise measurement of the perturbation vector (Mohapatra et al. 2012; Colebatch et al. 2016; Chen et al. 2017). In these latter cases, both magnitude and "onset" of the perturbation were quantified based on its acceleration effects on certain body parts, but such approach is limited by the dependence on the actual body part absorbing the perturbation.

<u>The linearity of the postural response</u>

In his early work on posture system identification Maki (Maki 1986; Maki and Fernie 1988) emphasized the opportunity to look for linear time-invariant models which, compared to nonlinear ones, offer computational simplicity and ease of interpretation while still proving an adequately realistic representation at least within a limited range of input amplitudes and time periods. When the human body is perturbed by an A-P perturbation a PR in the form of ankle strategy may be elicited (Shumway-Cook and Woollacott 2007) causing displacement of the CoP (Mohapatra and Aruin 2013). In fact, other postural strategies may come into play only when the base of support is of limited length (hip-strategy) or when the magnitude of perturbation increases (hip strategy, step strategy) (Horak and Nashner 1986; Horak et al. 1997). Whether or not the PR to forwardly imparted perturbations are substantially linear, or non-linear, may to a large extent depend on the magnitude of the perturbation. In a study where a posteriorly directing constant force of 10-40N against the sternum of a standing subject was unexpectedly withdrawn, leading to a compensatory forward body movement, the authors demonstrated that under such a 'regime of small oscillation' the PR could be described in terms of a linear 3-link inverted pendulum (Bortolami et al. 2003), while stronger perturbations could result in additional strategies and violation of linearity. Based on that model, they also postulated that the magnitude of PR would be proportional to the magnitude of the stimulus although that was not experimentally verified. With these considerations in mind, low-magnitude perturbations were delivered focusing on the CoP as the exclusive outcome measure in order to explore: 1. the identification of the best input variable, i.e., the perturbation parameter that better correlates with the CoP response; 2. the I/O characteristic of the system; 3. the reliability of this assessment. Within the context of the specific experimental set-up of this study, neither of these questions may have been addressed before.

A preliminary study on postural responses to perturbations applied to the high back (Kim et al. 2009) already evidenced a positive correlation between peak force and CoP displacement, which was normalized to foot length. In the present study, the intra-individual examination of postural responses evidenced a moderate correlation of the PR with PF of the perturbation but a much stronger correlation with the associated *I*. This proportionality between stimulus magnitude and postural response confirms and supports the concept of linear behavior of postural control (Bortolami et al. 2003; Kim et al. 2009). Moreover, the apparent superiority of I over PF in driving the CoP is a new finding that to the best of our knowledge has never before been reported. It is however in agreement with the following (simplified) biomechanical considerations. If, in the first instance, we approximate the body as a passive inverted pendulum with counteracting spring(s), its CoP displacement upon perturbation will depend on the acquired angular momentum which in turn is proportional to the imparted impulse, not to just to the force. In fact, two equal forces applied for different durations would result in different impulses, which would produce different changes in momentum and result in different CoP displacements. Of course, the overall response will also be affected by the active (reflex) reaction that will take place.

Given the good proportionality between *I* and ΔCoP, their ratio, ΔCoP_n, provides a synthetic descriptor of the individual postural performance. Surprisingly, this index differed quite considerably among different subjects (Fig. 6a) and was not correlated with their weight or height.

It may be observed that the correlation between ΔCoP and *I* was not high in all subjects (Fig. 5). While the fact that all perturbation were highly variable in terms of PF, duration and *I*, may partly account for the intra-subject variability, individual differences in, e.g., postural steadiness, degree of relaxation, internal motor setting, as well as the capacity to switch among different postural reaction strategy may account for the different response patterns exhibited by the different subjects. This issue was not here specifically investigated but could be more effectively addressed in future studies if other sources of variability are reduced, e.g., by delivering accurately controlled perturbations.

In a number of studies the ΔCoP response was normalized to the length of the foot or of the base of support (BoS) (Mille et al. 2003; Tortolero et al. 2006; Kim et al. 2009). While this approach appears appropriate when the perturbations push the CoP to the limits of the base of support (Mille et al. 2003), it may be of a lesser importance under the present test conditions where the perturbations were of low magnitude, driving the body forward to a limited extent and thus far from eliciting a stepping response. In fact, a linear behaviour of the system is expected in this working range while the limits of the base of support represent the dominant saturation-like non-linearities of the system (Maki and Fernie 1988). It is also possible that subtle changes in the muscular response pattern would have taken place among subjects of similar stature but different foot size but in this study monitoring of muscular activity was not incorporated.

Considering the differences in the experimental settings, the present results are in agreement with others. In a recent study, anterior and posterior perturbations were manually imparted to the shoulders of healthy subjects (Colebatch et al. 2016), resulting in an average CoP latency of 99 ms (measured with respect to body acceleration at C7) and CoP displacement in the range 44.7-80.4 mm, which was not correlated with height and weight of the subjects. Other studies (Mohapatra et al. 2012; Mohapatra and Aruin 2013) used the CoP to compare anticipatory and compensatory responses following a posteriorly-directed perturbation whose magnitude was equal to 5% of the subjects' bodyweight (about 30 N on average), obtained by releasing a pendulum at a distance of 60 cm from the body which hit the standing subjects at shoulder level. The resulting backward ΔCoP displacements ranged between 4 ± 0.3 and 6 ± 0.3 cm, in the different conditions (Mohapatra et al. 2012).

Reliability

Very few studies explored the reliability of perturbation-induced PR (Maaswinkel et al. 2016). Recently the issue was addressed by two studies in which the perturbation was provided by a moving platform. In one study, the reliability of 13 CoP variables was investigated in 10 different conditions, including one- or two-legged stance and antero-posterior perturbations with eyes open or closed and was reported to be generally poor and below acceptance levels (ICC = 0.70) (Robbins et al. 2017). However, two factors may account for this disappointing outcome: 1) the considered variables were not specifically designed to assess responses to perturbations and 2) the test-retest reliability was performed on average values obtained from only 3 PR, for each condition (Robbins et al. 2017). In another study, larger ICC values, generally above 0.7 and sometimes above 0.9, were calculated for intra- and inter-day reliability of CoP excursion tested on single-legged conditions, in response to antero-posterior and medio-lateral perturbations (Schmidt et al. 2015). In this case, measurements were performed over short time intervals of 0-70 and 70-260 ms from the start of perturbation, corresponding to the passive and active PR phases, respectively, and averages of 12 PR were considered.

The reliability of anteriorly-directed perturbations imparted to trunk of standing subjects has not been hitherto studied. The ICC values of 0.93 and 0.82 here reported for ΔCoP_n in response to inter-scapular and lumbar Ps, respectively are very high and in line with those reported by Schmidt et al. (2015). It should be observed that reliability, as assessed by ICC, increases with inter subject variability and with increasing number of trials in each condition (Maaswinkel et al. 2016). We here averaged a large number of trials (n=20) also due to the fact that the magnitude of stimuli was largely variable (20-60 N). Devices allowing to deliver perturbations with pre-set force or impulse are currently under development (Maffiodo et al. 2020). Whether this possibility will permit shortening of the test while preserving reliability will be investigated in future studies. In any case, the high reliability observed in the present conditions makes this approach worth of further consideration for the assessment of postural control, although investigations over larger population samples are required to clarify the actual variability across subjects, also depending on gender and age.

Inter-scapular vs. lumbar perturbations

Postural responses were distinctly and significantly affected by the site of perturbation presenting shorter latency and lower ΔCoP responses for lumbar perturbation.

The lower ΔCoP responses can be partly explained by biomechanical considerations (again referring to the inverted pendulum model): for the same magnitude of perturbation, the torque acting on the subject is smaller with lumbar than with interscapular perturbations due to the shorter force lever arm (distance between the point of application of the perturbation and the ankle), thus resulting in smaller angular momentum and smaller ΔCoP. Furthermore, other factors, such as differences in the PR synergies activated for the different stimuli (Horak et al. 1997; Rogers and Mille 2018) may also be involved and account for the difference in latency. While the present data do not allow for detailed analysis of the pathways underlying the PR it may be speculated that a perturbation delivered at a higher height would differently affect the differently body segments, possibly resulting in a more complex postural destabilization of the multi-joint inverted pendulum, requiring a more carefully coordinated and slower response. On the other hand, the individual ΔCoP_n assessed at inter-scapular and lumbar sites were highly correlated, suggesting that they actually estimate the same posture control feature. This incidentally means that if the ΔCoP_n becomes an applicable outcome measure only one of the two should serve as the perturbation site. However, given the superior reproducibility of its respective findings along with the comparative facility in imparting the perturbation, the inter-scapular site would be the most favourable one.

Functional significance

The present study indicates that ΔCoP and the latency are significantly and positively correlated which suggests that they might be also functionally linked. In order to maintain postural stability in response to the anterior perturbation the plantar flexion of the feet, which produces the increase in CoP has to counteract both the imparted angular momentum as well as the additional torque produced by the forward displacement of the centre of mass (Bortolami et al. 2003; Le Mouel and Brette 2017). Since the latter component increases with time as the body progressively leans forward, the longer the response latency, the larger will be the effort required, and consequently the extent of CoP displacement. Although several studies have investigated CoP responses to postural perturbations of different types, to our knowledge the correlation with the response latency has not been previously reported. It was however reported that a higher reaction time, used to simulate elderly people behaviour, contributes to reduce the maximal release angle sustainable with a single step (Vallée et al. 2015). It was also observed that “persons may compensate for delayed postural responses latencies by increasing the magnitude of their responses” (Horak et al. 1997). To what extent this inter-individual difference is due to a different motor setting (possibly related to the subjective understanding of the task), physical fitness, muscle stiffness or other anatomical/structural differences cannot be deduced from the present data and will need to be addressed in future investigations. We can however speculate that reduced latency and CoP displacement represents an advantage for postural control as it reduces the risk for CoP of exceeding the limits imposed by the base of support as well as the need to call for other balance control strategies. On this basis the ΔCoP_n can be considered as an index of effectiveness of the postural control, lower values indicating better control.

Limitations of the study

This preliminary study has a number of limitations. First is the small subject sample, consisting of young healthy men, and the perturbation type, limited to anterior perturbations, which do not permit generalizations beyond this group and methodology. Next is the absence of a kinematic element based on, e.g., accelerometers or simple optical sensors, to supplement the CoP data with information about displacement of the centre of mass. In this respect, one should consider the stabilization of the device and variability of direction of the force as contributing factors to the variability of the outcome measures.

However, such sensors depend on personal anthropometrics, cannot be firmly fixed to the body, are prone to external 'noise' and may not be essential in the framework of standard clinical testing. We believe that the force platform and its derived parameters may suffice for this purpose. Furthermore, some force platforms enable measurement of the footprint and therefore incorporation of a quantitative measure of the base of support. The absence of EMG records is more of a hindrance and we plan to incorporate such measurements in future studies. Finally, this preliminary study was designed to explore the input-output relationships of the system and although the approach seems to be valid, a proper reproducibility study needs to be conducted by carefully controlling the magnitude and duration of the perturbations. Such an analysis is critical for clinical applications where the setting of statistically based cut-off values and their employment in specific clinical disorders enables a decision as to whether a meaningful change in the said PR has taken place following intervention (Dvir 2015).

Finally, given the biomechanical nature and the specific objectives of the present investigation, no attention was devoted to the study of how anteriorly-directed perturbations altered the sensory-motor dynamics of quiet standing. The complex mechanistic principles underlying the compensatory responses to the externally imposed perturbations will have to be elucidated in future research.

Conclusions

A new approach to the investigation of the control of standing balance has been presented, in which uncontrolled but accurately measured postural perturbations were manually delivered. The preliminary results suggest that the impulse rather than the peak force better characterizes the perturbation and correlates with the postural response in terms of ΔCoP. On this basis a new postural index ΔCoP_n is proposed which exhibits good reliability, and which may indicate (with decreasing values) the effectiveness of the human body in counteracting challenges to balance maintenance. Further studies will be necessary to better understand its functional meaning and its possible clinical applications.

CONFLICT OF INTEREST

A patent application which includes part of the information presented in this article has been presented by some of the authors (ZD, CDB, DM, WF, CF and SR)

DATA AVAILABILITY

The datasets generated during and/or analysed during the current study are available from the corresponding author on reasonable request.

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FIGURE LEGENDS

Fig. 1 The manual perturbator. 1 – handle, 2 – load cell, 3- interface element with rubber foam lining

Fig. 2 The experimental Set Up. I.S.= inter-scapular and L=lumbar sites

Fig. 3 Representative recordings of force (upper window) and ensuing displacement of the Center of Pressure (CoP, lower window), relative to an inter-scapular perturbation in one subject. The displacement is positive in the forward direction.

ΔCoP: maximum displacement of CoP with respect to pre-stimulus value; Lat: latency of the CoP response; TtP: time-to-peak; D_CoP: Duration of CoP response.

Fig. 4 Scatter plots indicating the correlation between the center of pressure response (ΔCoP) and the magnitude of the perturbation expressed in terms of Peak Force (left) and Impulse (right) for both Inter-Scapular (top) and Lumbar (bottom) perturbations, in a representative subject. Each point refers to a single perturbation

Fig. 5 Distribution of the Pearson's Correlation Coefficients, for the ΔCoP – Impulse (light grey) and the ΔCoP - Peak Force (dark grey) correlation, for inter-scapular (a) and lumbar (b) perturbations. All box plots include data from the 10 subjects, a lower number of dots may appear due to overlapping values.

Fig 6 Individual normalized postural responses (ΔCoPn) scores in the two sessions for inter-scapular (a) and lumbar (b) perturbations, error bars representing the standard error of the mean. (c) Correlation between individual ΔCoPn scores achieved with inter-scapular and lumbar perturbations. (d) Correlation between ΔCoPn scores and Latency for inter-scapular (IS) and lumbar (L) perturbations.