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# Stepping boundary of external force-controlled perturbations of varying durations: Comparison of experimental data and model simulations

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ABSTRACT

This study investigated the stepping boundary – the force that can be resisted without stepping – for force-controlled perturbations of different durations. Twenty-two healthy young adults (19–37 years old) were instructed to try not to step in response to 86 different force/time combinations of forward waist-pulls. The forces at which 50% of subjects stepped ($F_{50}$) were identified for each tested perturbation durations. Results showed that $F_{50}$ decreased hyperbolically when the perturbation's duration increased and converged toward a constant value (about 10% BW) for longer perturbations (over 1500 ms). The effect of perturbation duration was critical for the shortest perturbations (less than 1 s).

In parallel, a simple function was proposed to estimate this stepping boundary. Considering the dynamics of a linear inverted pendulum + foot model and simple balance recovery reactions, we could express the maximum pulling force that can be withstood without stepping as a simple function of the perturbation duration. When used with values of the main model parameters determined experimentally, this function replicated adequately the experimental results.

This study demonstrates for the first time that perturbation duration has a major influence on the outcomes of compliant perturbations such as force-controlled pulls. The stepping boundary corresponds to a constant perturbation force-duration product and is largely explained by only two parameters: the reaction time and the displacement of the center of pressure within the functional base of support. Future work should investigate pathological populations and additional parameters characterizing the perturbation time-profile such as the time derivative of the perturbation.



## 1. Introduction

The neural control of human standing is concerned with keeping the body mass balanced above a base of support (BoS) provided by stationary feet. From a functional perspective, the feet-in-place responses provide only a weak capacity to restore balance when threatened by internal or external disturbances. Stepping or grabbing responses to instability reconfigure the BoS and provide a much more efficient solution to preserve balance and stop falling (Maki and McIlroy, 1997). These automatic change-in-support reactions play a more important functional role in maintaining equilibrium than feet-in-place responses. Contrary to traditional view, they are not just strategies of last resort but are often initiated before balance approaches instability, particularly for older people (Mille et al., 2003; Pai et al., 2000).

Balance and stepping research has often applied perturbations to the body that directly constrain the mechanical state of the body and thus that constrain subjects' responses. Examples are: (i) tether-release experiments (Carbonneau and Smeesters, 2014; Hsiao-Wecksler and Robinovitch, 2007; Thelen et al., 1997) where the initial lean angle and a null velocity are imposed, (ii) position-velocity controlled waist-pull experiments (Mille et al., 2003; Rogers et al., 2001) in which the pelvis is shifted forward at a specified amplitude and velocity whatever the subject's responses, or

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(iii) position-velocity controlled support surface translation (Maki and McIlroy, 1997; Pai et al., 2000) where the feet are moved relative to the center of mass (CoM) at specified amplitude and velocity. The common feature of these perturbations is that the body displacement imposed by the perturbation does not change according to the subject's response. These perturbations place a subject in a given perturbed state mechanically defined by the position and velocity of his CoM relative to the BoS. From this state, stepping boundaries (whether a feet-in-place response can restore balance or a step is needed) are determined by neuromuscular characteristics of the subjects and the direction(s) of perturbation. However, other characteristics of the perturbations, i.e. how the mechanical state at the end of the perturbation is reached, do not influence the outcome of such perturbations (Moglo and Smeesters, 2005; Vallée et al., 2015).

Few studies focused on compliant perturbations (i.e. perturbation during which a subject's response modifies the body displacement induced by the perturbation such as force-controlled perturbations or long-lasting platform perturbation) despite them being more common in daily-life: a gust of wind, push by another person, public transportation decelerations, etc. For these more natural perturbations, the mechanical state of the person is the result of both the perturbation and the resistance (passive + person's responses) to the perturbation. As such, the time-profile of the perturbation, and in particular its duration, might greatly influence its outcomes. To our knowledge, relatively few studies have investigated stepping boundary with compliant perturbations and these have been limited to a single duration of perturbation (Sturnieks et al., 2012, 2013). There is thus a need to understand stepping reaction to compliant perturbation of various durations.

The present study investigated the stepping boundary during forward force-controlled (i.e. compliant) perturbation of varying durations delivered at waist level and confronted the experimental results with a simple biomechanical model that could predict when a subject had to step. We expected an inverse relationship between the force and the duration of the perturbation: the longer the duration, the smaller the force required to trigger a step.

## 2. Method

### 2.1. Experimental data

#### 2.1.1. Subjects

Participants were twenty-two adults (5F, 17 M) aged 19–37 (mean 25.5 SD 4.13) years with mean height 174.3 cm (SD 7.14) and weight 69.9 kg (SD 10.2). Exclusion criteria were significant neurological (e.g. stroke, Parkinson's disease, neuropathy), musculoskeletal (e.g. joint replacement, leg or back pain), medical or balance disorders (e.g. cardiac, metabolic, respiratory, depression, surgery within 6 months) that could limit a person's movements. All participants gave written informed consent prior to the study, which was approved by the Institutional Review Board of the Institute of Movement Sciences, Aix-Marseille University and conducted in line with the principles of the Declaration of Helsinki.

#### 2.1.2. Protocol

Subjects stood on a force platform (OR6-6, AMTI, MA) that recorded the forces under the feet from which the position of the center of pressure (CoP) was calculated. They adopted a natural and comfortable foot position that was traced onto the floor to replicate initial position between trials. The perturbation force was delivered by a computer-controlled synchronous servomotor (AKM52M, Kollmorgen, VA) that pulled through a lightweight non-elastic Kevlar line to a firmly fitting belt around the subject's waist at upper pelvis level (Fig. 1A). A load cell (MLP100, Transducer Techniques, CA) coupled the cable to the belt to monitor the perturbation force. A baseline tension of 8 N kept the cable taught.

Body movements were recorded by a video motion analysis system (CodaMotion, Charnwood Dynamics, UK) with markers on the heels and over the C7 and S1 spinous processes. A real-time acquisition system (ADwin-Pro, Jäger, Germany) running at 10 kHz used customized software (Docometre) to control the force perturbations and acquire synchronous data. Force plate and load cell data were sampled at 1000 Hz and the video motion data at 100 Hz.

The test protocol began with 4 practice trials before commencing 86 different force-time combinations (15 forces between 40 and 180 N, and 9 durations between 150 and 3000 ms: Fig. 1B) in random order. Each trial lasted 5–7 s. Subjects were instructed to "*try not to step*" in response to the perturbations. Arm and other segmental movements were not constrained. The pull came at unexpected time (1–5 s) after a "*ready*" signal. The perturbation profile was a simple step reaching the target force and held for a prescribed time before release (Fig. 1A). The perturbation stopped prematurely only if the subject completed two steps (i.e. stepped off the force plate). If a step was not initiated, the subject could lean back to the initial position for the next trial. If they stepped, they repositioned to the set foot placement.

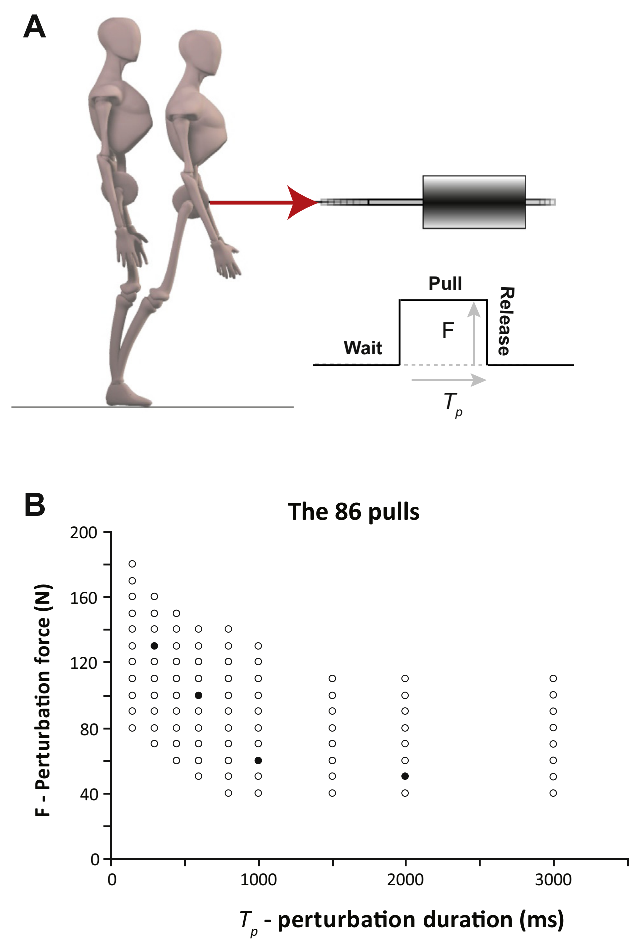


**Fig. 1.** *Experimental setup and protocol.* **A**: A rotary motor pulled subject forward by a flexible cable attached around his waist. The pulls started at an unexpected time and proceeded at test force ($F$) for a specific time ($T_p$) after which the cable tension was released and subject could lean back if a step had not already been initiated. **B**: Eighty-six pulls of different force ($F$) and duration ($T_p$) were delivered. Perturbations were presented in a random order different for each subject. Subjects started with 4 training trials (filled circles) to familiarize them with the perturbation.

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*2.1.3. Data analysis*

Each perturbation was described by its force normalized to subject weight: $\widehat{F}$ as a percentage of body weight (BW)) and its duration ($T_p$ in ms).

Subjects' responses were characterized by: (1) the presence or absence of a step, confirmed by vertical and anterior displacement of one heel marker, (2) the maximal antero-posterior CoP displacement after perturbation, which was just prior to toe-off if a recovery step was triggered ($CoP_{max}$ m), (3) the reaction time ($T_r$ ms) as time after force onset when CoP diverged more than 2SDs from baseline values (1 s before perturbation onset), and (4) the maximal trunk lean ($\theta_{max}$ in rad) as the angle from vertical to the line joining S1 and C7 markers.

For each subject and each duration for which it was possible, a force threshold was approximated as the mid-point between the largest force without a step and the smallest force with a step (Fig. 2). As we expected an inverse relationship between the force and the perturbation duration, the individual stepping boundary was described by fitting a hyperbolic function with a positive horizontal asymptote (Eq. (1)) using a linear least-square method. Thus, $\widehat{F}_{ind}$ represents the perturbation force required to initiate a step, constant $a$ defines the radius of curvature of the function and $c$ defines the horizontal asymptote, which describes the smallest force necessary to trigger a step - a force less than $c$ could be sustained indefinitely without stepping.

$$\widehat{F}_{ind} = \frac{a}{T_p} + c \tag{1}$$

As illustrated in Fig. 3, the probability of stepping for the entire group of participants was then calculated as a function of perturbation parameters from pooled individual force and duration data in two steps. First, for each perturbation duration ($T_p$), the step frequency ($f_{step}$) and perturbation force ($\widehat{F}$) data were used to identify the force at which all subjects stepped in 50% of presentations ($\widehat{F}_{50}$) by fitting the sigmoid function (Eq. (2)) using the Gauss-Newton non-linear least-mean-square method ($k$ is slope at the 50% point) (Table 1).

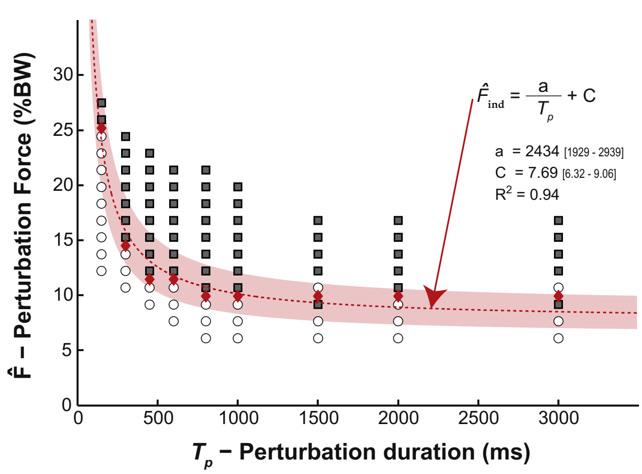


**Fig. 2.** *Individual stepping boundary.* This figure shows the responses obtained during a full set of trials for a representative subject. The demarcation between the force that causes subjects to step (squares) or not (circles) is obvious. For each duration for which it was possible, a force threshold was approximated (red diamond) as the mid-point between the largest force without a step and the smallest force with a step. A hyperbolic function was then fitted to these mid-points to describe the maximal force threshold of this subject as a function of the perturbation duration. Shaded areas around the threshold represent the 95% confidence interval for a and C. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

$$f_{step} = 1 - \left(1 + e^{(\widehat{F} - \widehat{F}_{50})/k}\right)^{-1} \tag{2}$$

Then, $\widehat{F}_{50}$ was expressed as a function of $T_p$ by fitting a hyperbolic function (Eq. (3)) similar to Eq. (1). Thus, $\widehat{F}_{50}^{hyp}$ represents the perturbation force that will initiate a step 50% of the time.

$$\widehat{F}_{50}^{hyp} = \frac{a}{T_p} + c \tag{3}$$

The quality of all fits was estimated by the coefficient of determination ($R^2$) and the root mean square value of the residuals (RMSE).

## 2.2. Balance recovery model

Based on previous modeling studies (Koolen et al., 2012; Vallée and Robert, 2015) we obtained a simple function to estimate if a recovery step is necessary for a given square force perturbation.

*2.2.1. Mechanical model and balance recovery reactions*

The human body was represented as a linear inverted pendulum and foot. Compared to the classical inverted pendulum model (e.g. Pai and Patton, 1997), the CoM remains at a constant altitude instead of rotating around the ankle, which proved to be a valid approximation for balance recovery movements (Hof et al., 2005; Pratt et al., 2006; Vallée et al., 2015). The maximum trunk lean angle was used to quantify hip rotation and thus hip strategy. Since its values were very small (see values of $\theta_{max}$ in Results), hip strategy was not included in the model.

The model is based on two main principles and on describing body state as a virtual point or *Extrapolated Center of Mass* (XCoM), which is the ground projection of the CoM augmented by a quantity proportional to its velocity. First, for a given body state in the absence of external perturbations other than gravity, standing balance can be maintained without stepping if the XCoM remains within the functional BoS (Hof et al., 2005; Pratt et al., 2006) - a reduction of the *anatomical* BoS considering neuromuscular constraints (King et al., 1994; Vallée et al., 2015). In this study, it boils down to the fact that the XCoM cannot move beyond $CoP_{max}$ which thus corresponds to the anterior edge of the functional BoS. Second, a pulling force can be withstood without stepping if standing balance can be maintained without stepping *at the end of the perturbation* (i.e. the previous principle applies).

Balance recovery responses were modeled by the displacement of the CoP within the BoS: during the initial period (i.e. between onset of the perturbation and $T_r$), the CoP remained at the balance point between the two ankles; at $T_r$ it instantaneously shifts forward at distance $CoP_{max}$ where it remained stationary. The subject's response is thus described in the model by two parameters only ($T_r$ and $CoP_{max}$).

*2.2.2. Maximum pull force tolerated without stepping*

The maximum force that can be withstood without stepping ($\widehat{F}_{max}$ - normalized by subject's weight) is that which brings the XCoM at $CoP_{max}$ at the end of the perturbation. It can be expressed as a function of perturbation duration and step reaction parameters (Eq. (4a) –see also details in Appendix).

$$\widehat{F}_{max} = \widehat{F}_{static} K_{T_r} f(T_p) \tag{4a}$$

$$\widehat{F}_{static} = \frac{CoP_{max}}{z_0} \tag{4b}$$

$$K_{T_r} = e^{-\omega_0 T_r} \text{ with } \omega_0 = \sqrt{\frac{g}{z_0}} \tag{4c}$$

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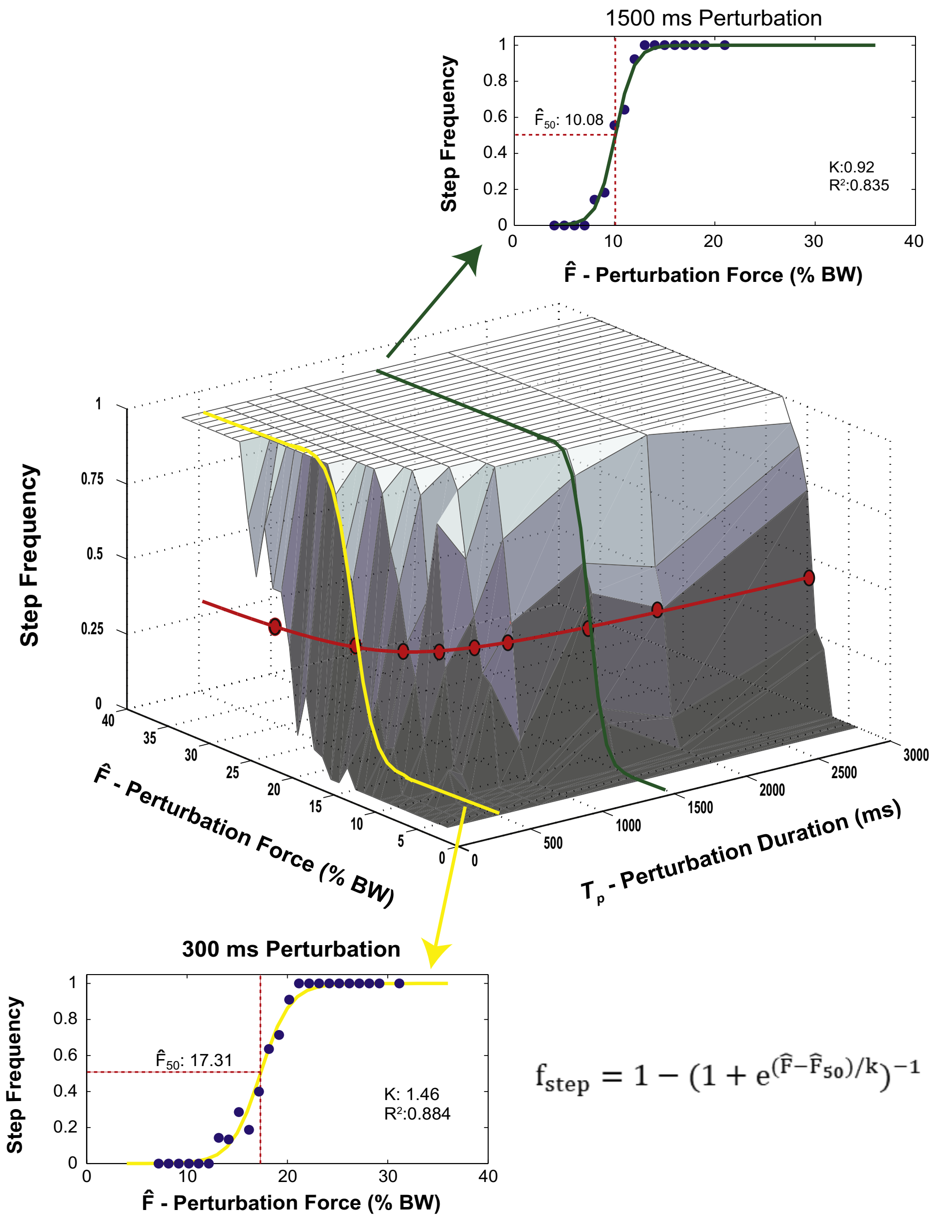


**Fig. 3.** *Group stepping frequency and boundary*. The relative step frequency (vertical) as a function of perturbation characteristics was calculated by summing individual data of normalized force and duration. For each perturbation duration, a sigmoidal function, shown for two of the perturbation durations (at the bottom for perturbation of 300 ms and 1500 ms at the top), was used to identify the force at which 50% of the subject stepped ($\hat{F}_{50}$). The solid horizontal curve is a hyperbolic function of duration fitted by a Gauss–Newton nonlinear least-mean-squares algorithm to the $\hat{F}_{50}$ data (circles). It represents force boundary at which the group steps in 50% of the pulls.

**Table 1**
Group force threshold.

| $T_p$ (ms) | $\hat{F}_{50}$ (%BW) | K |
|---|---|---|
| 150 | 24.4 | 2.1 |
| 300 | 17.3 | 1.5 |
| 450 | 14.2 | 1.3 |
| 600 | 12.4 | 0.9 |
| 800 | 11.6 | 1.2 |
| 1000 | 11.1 | 0.6 |
| 1500 | 10.1 | 0.9 |
| 2000 | 10.4 | 0.7 |
| 3000 | 9.5 | 0.8 |

$$f(T_p) = \frac{e^{\omega_0 T_p}}{e^{\omega_0 T_p} - 1} \tag{4d}$$

This force depends on 3 main factors. The first one, $\hat{F}_{static}$ (Eq. (4b)), is the maximum horizontal force at CoM height ($z_0$) that can be statically counterbalanced with the CoP at its maximal excursion ($CoP_{max}$) and normalized by subject's weight. The second factor, $K_{T_r}$ (Eq. (4c)), is a reduction coefficient expressing the fact that the delay in the reaction (with $T_r$ the reaction time) reduces the maximal force that can be applied (the longer $T_r$, the smaller $\hat{F}_{max}$). Finally, an exponentially decreasing function, $f(T_p)$ (Eq. (4d)) displays the influence of the perturbation's duration ($T_p$) on $\hat{F}_{max}$: the longer $T_p$, the smaller $\hat{F}_{max}$.

### *2.3. Comparison between predicted and experimental stepping threshold*

To compare the modeled $\hat{F}_{max}$ with the experimentally determined $\hat{F}_{50}$, Eq. (4) was used with the reaction parameters ($T_r$ and

**Table 2**
Mean and SD balance reaction characteristics of close-to-threshold trials.

| $T_p$ (ms) | $CoP_{max}$ (m) | | $T_r$ (ms) | | $\theta_{max}$ (rad) | |
|---|---|---|---|---|---|---|
| | Mean | *SD* | Mean | *SD* | Mean | *SD* |
| 150 | 0.161 | *0.019* | 102 | *21* | 0.34 | *0.15* |
| 300 | 0.160 | *0.015* | 116 | *19* | 0.38 | *0.17* |
| 450 | 0.155 | *0.017* | 117 | *16* | 0.41 | *0.14* |
| 600 | 0.153 | *0.017* | 123 | *20* | 0.37 | *0.17* |
| 800 | 0.153 | *0.012* | 114 | *20* | 0.38 | *0.18* |
| 1000 | 0.148 | *0.015* | 118 | *17* | 0.38 | *0.15* |
| 1500 | 0.146 | *0.015* | 115 | *23* | 0.33 | *0.13* |
| 2000 | 0.149 | *0.019* | 122 | *21* | 0.29 | *0.16* |
| 3000 | 0.150 | *0.027* | 116 | *25* | 0.36 | *0.18* |
| All | 0.153 | *0.018* | 116 | *20* | 0.36 | *0.16* |

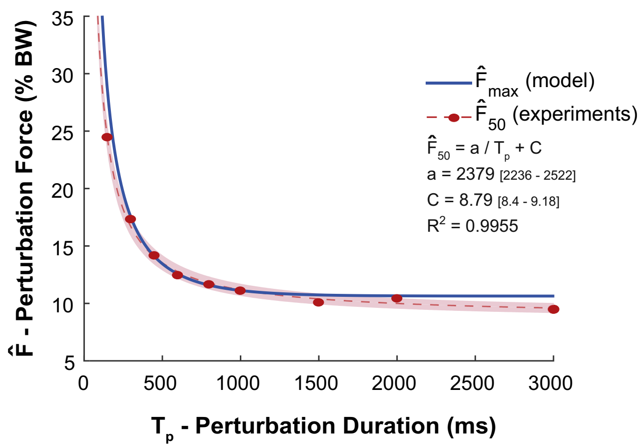


**Fig. 4.** *Comparison of stepping boundaries.* Comparison of the stepping boundary experimentally observed ($\hat{F}_{50}$, normalized force required to induce 50% of step – dashed red line) and the one estimated with the simple inverted-pendulum model ($\hat{F}_{max}$, maximal normalized force that can be withstand using reactions characteristics experimentally observed at $\hat{F}_{50}$– continuous blue line). The shaded areas represent the 95% confidence intervals. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

$CoP_{\max}$) determined experimentally for a perturbation intensity as close as possible to $\hat{F}_{50}$. As the specific $\hat{F}_{50}$ force was not tested directly, we interpolated between the perturbation forces either side of $\hat{F}_{50}$ (Table 2). CoM height ($z_0$) was derived from mean subject stature (Winter, 2009). Root mean square errors (RMSE) between $\hat{F}_{max}$ and $\hat{F}_{50}$ were then calculated.

## 3. Results

All subjects understood the task. One subject hopped in all trials and his data were not analyzed. With perturbations that did not trigger a step (i.e. that did not change the anterior limit of the BoS), subjects swayed around the ankles, sometimes rising on their toes, while trunk flexion at the hips was small.

As hypothesized, the stepping boundary decreased when perturbation duration increased (Fig. 4). This decrease was particularly marked for shortest durations (<1500 ms), while the boundary converged toward a horizontal asymptote for larger duration. The stepping boundary was highly consistent across subjects (Table 3) regarding in both its hyperbolic shape ($R^2$ consistently high) and the force values (similar fitting coefficients *a* and *c*). The group stepping boundary ($\hat{F}_{50}$) was also well approximated by the same simple hyperbolic function from Eq. (3): $R^2$ = 0.99 and RMS error (RMSE) of 0.30% BW. Fitting the exponential function from Eq. (4) yield to similar results (see Appendix).

**Table 3**
Parameters of the individual stepping boundary $\hat{F}_{ind}^{hyp} = \frac{a}{T_p} + c$.

| Subjects | a | 95% CI | | C | 95% CI | | $R^2$ |
|---|---|---|---|---|---|---|---|
| 1 | 2416 | 2069 | 2762 | 8.04 | 7.09 | 8.98 | 0.97 |
| 2 | 2477 | 1858 | 3097 | 10.15 | 9.12 | 11.18 | 0.93 |
| 3 | 3641 | 2876 | 4406 | 9.51 | 8.23 | 10.78 | 0.95 |
| 4 | 2402 | 1801 | 3003 | 10.61 | 8.97 | 12.24 | 0.92 |
| 5 | 1676 | 1379 | 1973 | 9.41 | 8.60 | 10.22 | 0.96 |
| 6 | 2294 | 1799 | 2789 | 10.25 | 8.91 | 11.60 | 0.94 |
| 7 | 2230 | 1414 | 3046 | 8.94 | 7.58 | 10.30 | 0.86 |
| 8 | 2468 | 1780 | 3157 | 10.71 | 9.57 | 11.86 | 0.92 |
| 9 | 2816 | 2163 | 3469 | 8.42 | 6.64 | 10.19 | 0.93 |
| 10 | 2805 | 2463 | 3147 | 9.35 | 8.42 | 10.28 | 0.98 |
| 11 | 1978 | 1535 | 2421 | 8.56 | 7.35 | 9.76 | 0.93 |
| 12 | 1495 | 1117 | 1872 | 8.20 | 7.18 | 9.23 | 0.92 |
| 13 | 2452 | 1789 | 3115 | 9.84 | 8.74 | 10.95 | 0.92 |
| 14 | 2434 | 1929 | 2939 | 7.69 | 6.32 | 9.06 | 0.94 |
| 15 | 2447 | 1808 | 3087 | 8.52 | 6.78 | 10.26 | 0.91 |
| 16 | 2700 | 1892 | 3507 | 8.94 | 7.60 | 10.29 | 0.90 |
| 17 | 2325 | 1544 | 3106 | 9.35 | 7.22 | 11.47 | 0.86 |
| 18 | 1893 | 1377 | 2408 | 8.54 | 7.14 | 9.94 | 0.90 |
| 19 | 2507 | 2314 | 2701 | 11.29 | 10.76 | 11.81 | 0.99 |
| 20 | 2133 | 1579 | 2686 | 9.46 | 7.96 | 10.97 | 0.91 |
| 21 | 1966 | 1483 | 2448 | 9.64 | 8.33 | 10.96 | 0.92 |
| **M** | **2359.71** | | | **9.31** | | | **0.927** |
| **SD** | **449.75** | | | **0.95** | | | **0.03** |

Table 2 presents the group mean balance response parameters of the close-to-threshold trials (mean of the two perturbations either side of $\hat{F}_{50}$, see method) for each perturbation duration. Even for these close-to-threshold perturbations, trunk rotation ($\theta_{max}$) remained limited, which justifies our modeling hypothesis that neglects the hip strategy. Furthermore, balance reaction parameters were uniform across subjects and duration of perturbations (see the small values of SD in Table 2). The model was thus parametrized using uniquely values of $T_r$ and $CoP_{max}$ averaged across subjects and perturbation durations.

Using Eq. (4), we estimated the maximal force that can be withstood by an average subject (see Method). This $\hat{F}_{max}$ threshold computed from the biomechanical model matched well the experimental $\hat{F}_{50}$ data (see Fig. 4) with small residuals (RMSE = 1.40% BW). This indicated that the stepping boundary was largely explained by the inverted pendulum model.

## 4. Discussion

This study measured for the first time the stepping boundary at which forward force-controlled (i.e. compliant) perturbations of different duration trigger a step and identified a simple biomechanical model that accurately predicts and explains this sagittal boundary by only two parameters: the reaction time and the displacement of the CoP within functional BoS.

### *4.1. Characteristics of the stepping boundary*

As hypothesized, the force required to trigger a step is strongly dependent of the duration of the perturbation: it decreased when the perturbation duration increased. More precisely, this stepping boundary is reliably described by a hyperbolic function ($\hat{F}_{50}^{hyp}$ Eq. (3)) in the force-duration space. This is true for the individual stepping boundary as well as for this homogenous group, suggesting that deviations from normative values could be use in identifying particular abnormalities.

The vertical asymptote is at duration zero, predicting that as duration approaches zero, increasingly larger forces will trigger a step, although at some point this is no longer achievable experimentally. For a 100 ms perturbation, the force needed to

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trigger a step is approximately 25% BW. As participants rated the maximum force used in this experiment (180 N) as "rather violent," it is not practical to test higher force levels experimentally.

The horizontal asymptote (coefficient $c$ in Eqs. (1) and (3)) represents the maximum force that can be resisted indefinitely without triggering a step. For young subjects, this force corresponded to 8.8% BW. It is likely that for very long-lasting perturbations this force will decrease further due to neuromuscular fatigue. Indeed, the ankle torque used to maintain a normal standing position is around 50 Nm for a typical person (70 kg with 1 m CoM height leaning 4° forward). This corresponds to 15–20% of the contractile strength of the soleus muscle, the most active muscle during quiet standing (Joseph and Nightingale, 1952; Morin and Portnoy, 1956). Adding a perturbation force of 9% BW will more than double ankle torque (increase of 61 Nm for the 70 kg person). Effects of muscular fatigue over long time courses will probably decrease available strength so that stepping would be induced at lower perturbation force.

Previous studies from Sturnieks et al. (2012, 2013), also investigated stepping boundaries for a single duration of force-controlled perturbation. The maximal 600 ms trapezoidal forward force that young subject could withstand without stepping was about 81 N, (i.e. about 11.8% BW considering an average BW of 70 kg) corresponding closely with the current study ($\widehat{F}_{50}$ = 12.4% BW for $T_p$ = 600 ms).

This study shows that for compliant (force-controlled) perturbations, the stepping boundary is strongly affected by the perturbation duration. This is particularly true for shortest durations (<1500 ms) that are the most commonly encountered in daily life. This duration effect should thus be considered when evaluating the risks associated with compliant perturbations and/or when comparing compliant perturbation studies. Still, further investigations are necessary: (1) to extend these results to other populations (although a similar effect is expected), and (2) to investigate other perturbation's time profile parameters such as the time-derivative of the perturbation (Graaf and Van Weperen, 1997; Vallée, 2015; Vallée et al., 2016).

### 4.2. Modeling approach

Response to the perturbation was characterized by three parameters: the reaction time ($T_r$), the maximal displacement of the CoP within the BoS ($CoP_{max}$) and the maximal trunk lean angle ($\theta_{max}$). Interestingly, values of these parameters for the close-to-threshold trials were almost invariant across subjects and duration of perturbations, i.e. all subjects of this homogeneous group of population used similar reactions when pushed to their limits, independently of the duration of perturbation.

The stepping boundary in force-duration space is well predicted using a simple biomechanical model. The predicted force-duration relation (Eq. (4a)) and the experimental data display similar shape and properties: an inverse exponential decrease that converges toward a horizontal asymptote. Likewise, the predicted stepping boundary $\widehat{F}_{max}$ obtained by inserting experimental values of the main model parameters approximated to the experimental observations $\widehat{F}_{50}$ (see Fig. 4).

This model shows that: (1) the overall influence of the perturbation duration on the stepping boundary, represented by the factor $f(T_p)$ in Eq. (4a), is independent of subject's neuromuscular capacities; (2) for a given duration of perturbation, the maximum force that can be resisted without stepping depends primarily on subject's capacity to generate enough ankle torque quickly to negate the perturbation. The model represents these characteristics as the reaction time ($T_r$) and the maximum forward distance at which the CoP can be instantaneously shifted and held ($CoP_{max}$). Thus, it could be applied to assess the consequences of potential degradation of these physical capacities. Fig. 5 displays the variation of $\widehat{F}_{max}$ as a function of $CoP_{max}$ and $T_r$ for two different perturbation durations: limiting $CoP_{max}$ or increasing $T_r$ by two reduced $\widehat{F}_{max}$ by about 50% and 30%, respectively. One can also note that the maximum force is affected more, in net values, with shorter perturbation durations.

Obviously, one should be cautious when trying to use this model to represent non-tested population behavior. In particular, the very simple description of the recovery strategies performance using only $T_r$ and $CoP_{max}$ might not be sufficient to capture the behavior of elderly or pathological subjects. This model could still be refined. Biomechanically, arm swing and counter movements are not included but these could help counteract destabilizing forces and potentially avoid stepping, thereby affecting the estimate of the stepping boundary. In determining the stepping boundary, we implicitly assume that the most efficient reactions are used systematically however this might not be adapted to study bellow threshold perturbations effect or difficulty (Horak, 2006; Pai et al., 2000; Rogers et al., 2001). Adaptation and learning with changing stepping strategies is not considered.

### 4.3. A constant impulse triggers a step

This model is based solely on biomechanical principles. While it identifies the stepping boundary, it provides no insight into sensory information used by the subjects to trigger, or not trigger, a step. The stepping boundary, with $\widehat{F}_{50}$ well catch by a simple hyperbolic curve described by Eq. (3) suggests a possible hypothesis. Rewriting this equation as:

$$\widehat{F}_{50} \approx \frac{a}{T_p} + c \Longleftrightarrow (\widehat{F}_{50} - c)T_p \approx a \tag{5}$$

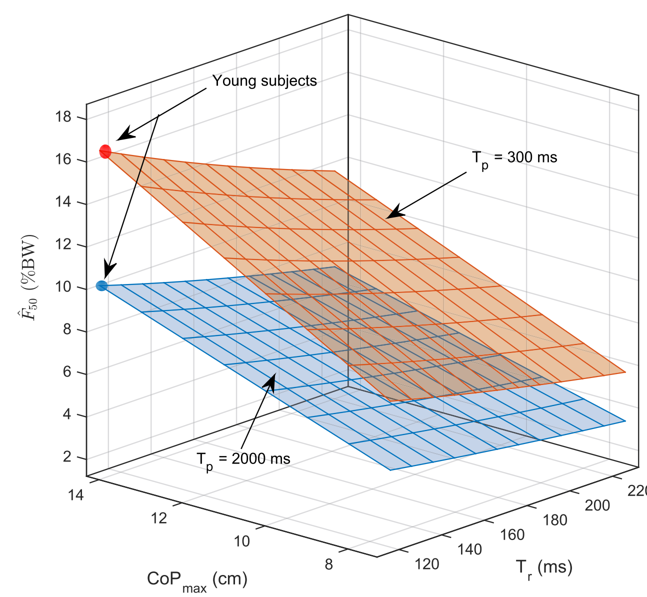


**Fig. 5.** *Effect of varying reaction characteristics.* $\widehat{F}_{max}$ was computed from Eq. (4) for two perturbation duration $T_p$ (300 ms in red and 2000 ms in blue) and reaction time ($T_r$) and maximal CoP excursion ($CoP_{max}$) values ranging from their mean experimental value (red and blue dots) to the double or half, respectively. It is clear that limiting $CoP_{max}$ (i.e. reducing the functional BoS by half) or doubling $T_r$ reduced the maximum force to trigger a step by about 50% and 30%, respectively. One can also note that this force is more affected, in net values, for shorter perturbation durations. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

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shows that the stepping boundary corresponds to a constant impulse of force (constant force-duration product). From Newton's second law, the impulse of force is the change in momentum of the body (mass $\times$ $\Delta$velocity). Thus, the trigger for a step could be a threshold velocity change, which could be detected by proprioceptive or vestibular afferent inputs (Fitzpatrick and McCloskey, 1994). Further experimentations are necessary to test this hypothesis.

In conclusion, this study has demonstrated that perturbation duration has a major influence on the balance responses to compliant perturbations, such as the force-controlled pulls used here. The stepping boundary is described by a constant perturbation force-duration product and is largely explained by only two parameters: the reaction time and the displacement of the CoP within the BoS. Future work could investigate pathological populations and additional parameters characterizing the perturbation time-profile such as the time derivative of the perturbation.

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## Conflict of interest statement

Authors have no conflict of interest to report in this research.

## Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at https://doi.org/10.1016/j.jbiomech.2018.05.010.

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