Predicting Multiple Step Placements For Human Balance Recovery Tasks

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Abstract

Stepping is one of the predominant strategies to restore balance against an external perturbation. Although models have been proposed to estimate the recovery step placement for a given perturbation, they suffer from major limitations (step execution time usually neglected, no more than a single step recovery considered, etc.). The purpose of this study is to overcome these limitations and to develop a simple balance recovery model which can predict a complete multiple step recovery response. Inspired by the field of walking robots, we adapted a control scheme formerly proposed for biped robot locomotion. The scheme relies on a Linear Model Predictive Controller (LMPC) which estimates the best foot placements to zero the velocity of the Center of Mass (CoM), i.e. to reach a steady posture. The predicted step placements were compared against previously reported experimental data for tether-release conditions. They match correctly for various perturbation levels and both single step or multiple steps recovery. Although the current model still suffers from limitations (e.g. limited to the sagittal plane), these results demonstrate its ability to reproduce balance recovery reactions for different experimental scenarios.

Keywords:
Balance recovery, Stepping, Model Predictive Control, Fall

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1. Introduction

The human body is an intrinsically unstable mechanical system. Yet, humans are able to routinely perform highly challenging activities in terms of stability such as standing, walking, running, or recovering from unexpected disturbances. This ability has been the focus of many research. In particular, there is an abundant literature on static or slightly disturbed balance, and some models have been proposed to reproduce human-like behavior in quiet stance (Bottaro et al., 2008; van der Kooij et al., 1999; Peterka, 2002; Qu et al., 2007).

On the other hand, the study of balance recovery under larger disturbances has been much less explored. It has been primarily investigated experimentally, by subjecting volunteers to a balance recovery task by various means such as sudden release from initial lean positions (Cyr and Smeesters, 2009; Do et al., 1982; Hsiao and Robinovitch, 1999), slips (Bhatt et al., 2005; Pai et al., 2003) platform translations (de Graaf and van Weperen, 1997; McIlroy and Maki, 1996; Robert et al., 2007), etc. The analysis of the recorded reactions has revealed different balance recovery strategies, usually classified into two groups: ankle and hip strategies (Horak and Nashner, 1986; Nashner and McCollum, 1985; Runge et al., 1999) and change of support strategies (Maki and McIlroy, 1997). It is now well known that this latter, and in particular single or multiple recovery steps, is the predominant strategy in trials where no instructions are given to the subjects about the recovery behavior (Maki and McIlroy, 1997) and among the elderly (McIlroy and Maki, 1996).

Prediction models have been proposed to forecast the consequence of a given balance perturbation in terms of recovery steps. Different studies proposed thresholds for the position and velocity of the Centre of Mass (CoM) relative to the support base beyond which balance cannot be retained without stepping. They are based on experimental observations (Mille et al., 2003; Smeesters, 2009) or simplified representations of the human body (Hof et al., 2005; Pai and Patton, 1997). This latter approach was also used to predict the shortest instantaneous step that would restore balance from a disturbed state (Millard
et al., 2009; Pratt et al., 2006; Wight et al., 2008; Wu et al., 2007). However, these formulations suffer from two major discrepancies: 1) No more than a single step recovery is considered; 2) The step execution time is neglected. Both of these assumptions are violated in reality. Hence, there is still a need for a more elaborate balance recovery model allowing for a correct prediction of the recovery behaviors in humans.

Developments in the field of control of biped robots could provide interesting ideas for multi-step balance recovery predictions. In particular, Herdt et al. (2010) recently proposed a scheme to estimate on-line the best step placement in order to follow a given reference value of the CoM velocity while preserving balance. It relies on a model predictive controller, i.e. the prediction is made over a time horizon and integrates the system’s evolution during this horizon. This scheme has proved to efficiently ensure a stable bipedal gait (i.e. multiple steps), while being relatively simple and computationally efficient. It thus appears as a promising formulation to overcome the two major limitations of the current balance recovery models.

The goal of this study is to adapt the scheme proposed by Herdt et al. (2010) for biped robot walking situations to human balance recovery tasks. The resulting balance recovery model will be used to predict the step lengths for different experimental data sets from the literature by employing simple models of human-body and inputting the step timings.

2. Methods

2.1. Model Description

In this study, the balance recovery is considered only in the sagittal plane using a mechanical model of the human body placed in closed loop with a controller. This mechanical model is an inverted-pendulum-plus-foot model representing the support limb (see Figure 1a). The trailing limb is not explicitly modelled and its influence on the system dynamics is neglected. The length of the pendulum is constant for each step but can change from one step to another.
The resulting trajectories of the Center of Mass (CoM) are thus circles of possibly different diameters (see Figure 5) and can experience only instantaneous double support phases.

Table 1 and Figure 1 about here

The feedback loop is based on a Model Predictive Control (MPC) approach (Figure 2), using an internal model which can be different and simpler than the real mechanical model. This internal model is considered for predicting future control actions over a given time horizon. Expectations about the evolution of the state of the system are expressed as a cost function. The adequate control actions are then selected by minimizing this cost function, given the current state of the system. The controller implemented in this study is largely based on the Linear MPC (LMPC) recently proposed by Herdt et al. (2010).

Figure 2 about here

The internal model used here is a classical Linear Inverted Pendulum (LIP) model (Kajita and Tani, 1991), consisting of two massless feet and legs, the whole body mass being concentrated at its CoM. The height of the CoM is considered constant and the contact forces below the feet are reduced to a single force acting at the Center of Pressure (CoP) (Figure 1b). The resulting dynamic equation can be written then as

\[
\ddot{x}_{\text{com}} = \frac{g}{h}(x_{\text{com}} - x_{\text{cop}}),
\]

where \( g \) is the norm of gravity force, \( h \) is the height of CoM, and \( x_{\text{com}} \) and \( x_{\text{cop}} \) are respectively the coordinates of the CoM and CoP along the X axis (see Figure 1b).

The timing of the foot contacts (take-off and landing) is fixed in advance. It consists of:

– a reaction time, \( T_{\text{reac}} \), between the onset of the perturbation and the beginning of the reaction (activation of the controller),
– an additional delay, the step preparation time $T_{prep}$, considered before the
initiation of the first step,

– the durations of the further steps (delay between contralateral feet landings)
defined by the values of $T_{step}$.

Future actions are predicted over a time horizon of duration $T_{horizon} = 1$ second,
over which the constraints on the CoP with respect to the positions of the feet
on the ground are checked every $T_{sampling} = 25$ ms. Incidentally, double support
phases are assumed to coincide with such sampling intervals, and last therefore
a mere $25$ ms, staying close to the instantaneous double support phases of the
mechanical model. The total number $m$ of steps considered within the time
horizon is determined by these different timings of foot contacts.

The cost function describing our expectations on the motion of the system
is mainly based on the regulation of the horizontal velocity of the CoM to a
reference value. In this study the reference value is set to zero: the subject tries
to maintain a steady state posture. The jerk of the CoM and the deviation of
the CoP from ankle positions are also introduced with a small relative weight
in order to generate smooth motions and a comfortable final posture. It results
in:

$$\min_u \| \dot{X}_{com} \|^2 + w_1^2 \| \ddot{X}_{com} \|^2 + w_2^2 \| X_{cop} - X_{ankle} \|^2$$  \hspace{1cm} (2)

with

$$u = \begin{bmatrix} X_f & \dot{X}_{com} \end{bmatrix}$$  \hspace{1cm} (3)

where

– $X_f$ is a vector of $m$ different positions that $X_{ankle}$ takes successively during
  the $m$ steps that happen during the prediction horizon,

– $\ddot{X}_{com}$ and $\dot{X}_{com}$ are vectors of $N = T_{horizon}/T_{sampling}$ values of the jerk and
  velocity of the CoM along the X axis at each sampling period during the
  prediction horizon,

– $X_{cop}$ and $X_{ankle}$ are vectors of $N$ positions of the CoP and support ankle on
  the ground along the X axis at each sampling period during the prediction
horizon,

- $w_1$ and $w_2$ are relative weight coefficients.

In addition, following constraints are applied on the system: i) the CoP has to remain within the BoS; ii) the step length is limited by considering a fixed maximal velocity of the swinging foot. Both constraints are linear.

Consequently, the determination of the optimal foot placements and CoM trajectory boils down to minimizing a quadratic cost function under linear constraints, i.e. to solve a Quadratic Program (QP). Details can be found in Herdt et al. (2010).

Once this optimal control strategy has been determined, it is applied to the mechanical model of the human body during a whole sampling period $T_{sampling}$, after which the state of the system (altitude and horizontal position, velocity and acceleration of the CoM) is measured again and the whole MPC is recomputed on a new time horizon shifted accordingly. The overall simulation time is set to 2 s.

2.2. Comparison with Experimental Data

In order to compare our prediction model with experimental data, we need experimental situations which: 1) involve single and multiple steps recovery strategies; 2) involve relatively simple perturbations in order to avoid complex perception models at this stage; 3) provide sufficient details about the perturbation, the delays and the resulting recovery steps. We used data from two studies that comply with these requirements. They focus on tether-release conditions, a common experimental protocol in which the subjects are inclined to a stationary forward leaning position thanks to a tether and a safety harness (c.f. Figure 1) and released in this unstable posture. These studies are:

- the study from Hsiao-Wecksler and Robinovitch (2007) reports the maximum inclination angles for single-step recovery given a specific step length. Young subjects were inclined forward and asked to recover balance after release by taking a single step, no larger than a given target length. The
maximum lean angle for four target lengths, averaged across subjects, were
used as inputs to our model. The predicted step lengths are compared to
the target lengths. For clarity in final results, step lengths expressed in
% of body height are transformed into meters, based on the average body
height of the subjects.

– the study from Cyr and Smeesters (2009) is to our knowledge the only
study about the threshold of multiple steps balance recovery in which
kinematics and timing of all recovery steps are reported. In this case,
the subjects were asked to recover balance without any restriction on the
maximum number of steps. The lean angle was gradually increased until
the subject failed to recover balance. For comparison, the reported average
maximum lean angle value of 30.7° is used as input to our simulation.

2.3. Selection of the Model Parameters

The first group of parameters describes the experimental scenario under
consideration. Values used for each simulated scenario are reported in Table
2. Subjects’ stature are used to adjust the dimensions of the mechanical model
(see Table 1 from (Winter, 1990)). The step timing are fixed based on the
experimental values reported and rounded-off to the nearest multiple of the
25 ms sampling period. While Cyr and Smeesters (2009) reported all values,
Hsiao-Wecksler and Robinovitch (2007) only reported the first step landing time
\( T_{\text{land}} = T_{\text{reac}} + T_{\text{prep}} + T_{\text{step}} \). However, earlier reports suggested that \( T_{\text{reac}} \) and
\( T_{\text{prep}} \) can be considered constant for these types of perturbations (Do et al.,
1982; King et al., 2005). They are thus fixed respectively to 75 ms and 150 ms
and \( T_{\text{step}} \) is defined as the difference between the reported \( T_{\text{land}} \) and the sum
\( T_{\text{reac}} + T_{\text{prep}} \).

Remaining parameters are related to the controller, do not depend on the
scenario and are reported in Table 3. The maximal velocity of the swinging foot
is set to 6 m.s\(^{-1}\), i.e. slightly faster than during normal gait (Winter, 1992).
The time horizon (1s) is chosen such that important events related to balance
recovery, such as stepping, could arise during this time. The simulation time
(2s) is chosen large enough to completely converge the CoM velocity to zero. In this first approach we consider an almost continuous control, i.e. the controller is called every sampling time (every 25 ms). The values of weight coefficients $w_1$ and $w_2$ are kept small ($10^{-2}$ each), their purpose being to smoothen the contact forces and a comfortable final posture as noted in Herdt et al. (2010). To analyze the effect of varying these weight coefficients on the final prediction of step length, a brief study is conducted for the four balance recovery scenarii of Hsiao-Wecksler and Robinovitch (2007) presented in the next section.

Table 2 and 3 about here

3. Results

3.1. Effect of the Relative Weight Coefficients

Table 4 shows the effect of varying the weight coefficients values $w_1$ and $w_2$ on the predicted step length in the four balance recovery scenarii from Hsiao-Wecksler and Robinovitch (2007). The parameter $w_2$ penalizing the divergence of CoP from the ankle position has almost no effect on the final choice of step length (c.f. first 2 rows of Table 4) as this objective could be achieved once the balance has been recovered. On the other hand, penalizing the jerk of CoM ($w_1$) has a small but noticeable influence on the step length (up to 15 % when $w_1$ ranges between $10^{-4} s^2$ and $10^{-1} s^2$, see Table 4). However, just like $w_2$, extreme values of $w_1$ either produce abrupt motion of CoP during the course of the motion ($w_1 = 10^{-4} s^2$) or result in unnecessary velocity fluctuations and longer recovery times ($w_1 = 10^{-1} s^2$). We therefore consider a middle value of $10^{-2} s^2$ for this study.

Table 4 about here

3.2. Comparison between experimental and simulated results

Comparison between the experimental and simulated results are displayed in Figures 3 and 4. White plots show the experimental results (average ± s.d.) while the black plots represent the corresponding model predictions.
Figure 3 shows the results for the 4 inclination cases considered by Hsiao-Wecksler and Robinovitch (2007). It can be perceived that simulated and experimental step lengths match well, in particular for the smaller inclination angles.

Figure 4 shows the results for the extreme inclination case of Cyr and Smeesters (2009) with no limit on number of steps. Stride lengths instead of step lengths are reported to be coherent with this study. Predicted and reported stride length are of the same order. Note that the third step, reported by Cyr and Smeesters (2009) but not predicted by the model, was only observed for two out of 28 subjects.

**Figure 3 and 4 about here**

Figure 5 shows the evolution of the mechanical model during the predicted recovery for the scenario of Cyr and Smeesters (2009). It can be seen that the CoM trajectory follows circular arcs of varying lengths.

**Figure 5 about here**

### 4. Discussion

The goal of this study was to adapt a control scheme initially proposed for the locomotion of biped robots (Herdt et al., 2010) and test it against experimental balance recovery data reported in the biomechanics literature. By only inputting the subject anthropometry and timing of the foot contacts, the recovery step placements are predicted with a reasonable accuracy for different scenarios: single and shortest step recovery for different perturbation levels (lean angles) up to the maximum perturbation recoverable in one step, and multiple step recovery for the maximum recoverable perturbation.

Two different human body models are used in this simulation: a mechanical model is used to represent the actual human body behavior, driven by a controller which used another internal model to anticipate the effects of control actions.
The principal interest of the \textit{internal} model is its simple linear dynamics (1), thanks to which the determination of the optimal control action boils down to solve a Quadratic Program, i.e. very fast and reliable computations.

At this stage, the \textit{mechanical} model is kept as simple as possible while still representing a correct overall kinematics for the tether-release scenario simulated. One limitation though regards the constant leg length \textit{during} each step. Although this hypothesis seems almost verified in experiments (for example see Figure 2 of the original article from Cyr and Smeesters (2009)), it prevents both the use of the CoM altitude as a control variable for the recovery and the possibility to generate or dissipate energy in the leg during steps. Moreover it is not compatible with double supports (the main reason for keeping the duration of double support extremely short in this study). Introducing compliance in the support leg (Geyer et al., 2006) seems a promising approach to overcome this limitation (Mordatch et al., 2010).

The predicted step lengths are coherent with the experimental observations, but still slightly larger than the experimental values in all cases (see Figures 3 and 4). A closer look reveals that these differences scaled roughly with the inclination level. This could be attributed to the fact that rotation of the upper body is not considered in the current model. Indeed, it is known that upper body inertia plays a role in the balance recovery (e.g. van der Burg et al., 2007; Horak and Nashner, 1986; McIlroy and Maki, 1994), and can be used to limit the recovery step length (Pratt et al., 2006). Moreover, Park et al. (2004) showed that upper body rotations scale positively with the level of perturbation. Taken together, these results tend to support our hypothesis: neglecting upper body rotations in the prediction model leads to over-estimate the recovery step length, and this bias increases with the perturbation level. Although this effect remains relatively limited, further developments should focus on introducing the upper body rotations in this model.

The criterion used in the controller to select the optimal recovery actions (CoM motion and step placements) is primarily based on the minimization of the CoM horizontal velocity at each instant of the time horizon. It thus tends to
produce maximal recovery performances. This corresponds to the experimental studies considered, which focused on maximal recoverable perturbations for different stepping constraints. Moreover, this choice allows to study extreme situations (e.g. what is the minimal distance to recover from a given perturbation), which are usually of primary interest. However, this criterion will likely have to be adapted, typically by introducing energetic cost, in order to represent sub-maximal balance recovery behaviors.

A further limitation of our scheme is the pre-allocation of the step timings. For the purpose of comparison in this article, we chose these step durations to correspond to the reported experimental values. However, more work is required to optimally adjust this duration within the optimization routine, resulting in a unified prediction balance recovery model for different perturbations.

Despite these limitations, the current scheme demonstrated its ability to reproduce balance recovery reactions for different experimental scenarios. The two major strengths making this scheme unique are: 1) the possibility to represent multiple steps recoveries, 2) the consideration of the system’s dynamics, and thus its evolution, all along the recovery. This work could further be extended to different kinds of perturbations and populations. Particularly, the simulation of an elderly reaction could be carried out by identification and modification of relevant morphological and temporal parameters (such as introducing reduced functional base of support as done by Pai and Patton (1997), or including larger reaction times reported by several studies (e.g. see Hsiao-Weckskler (2008))). In addition, future works will focus on including the sensory and cognitive aspects in the model as well as improvements in the mechanical model.

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Conflict of interest statement

Neither author has any conflict of interest to report in this research.
References


McIlroy, W., Maki, B., 1994. Compensatory arm movements evoked by transient perturbations of upright stance, in: Taguchi, K., Igarashi, M., Mori, S. (Eds.),


Table 1: Anthropomorphic proportions used in the simulation (see also Figure 1).

<table>
<thead>
<tr>
<th>Variable</th>
<th>Scaling</th>
</tr>
</thead>
<tbody>
<tr>
<td>Body Height</td>
<td>$H$</td>
</tr>
<tr>
<td>Initial pendulum length, $l$</td>
<td>$0.575 \times H$</td>
</tr>
<tr>
<td>Foot length, $l_f$</td>
<td>$0.152 \times H$</td>
</tr>
<tr>
<td>Horizontal ankle-to-heel distance, $a$</td>
<td>$0.19 \times l_f$</td>
</tr>
<tr>
<td>Ankle height</td>
<td>$0.039 \times H$</td>
</tr>
</tbody>
</table>
Table 2: Simulation parameters used as input for the five simulated scenarios: lean angles ($\theta$), body height ($H$) reaction times ($T_{\text{reac}}$), step preparation times ($T_{\text{prep}}$) and leg swing times ($T_{\text{step}}$).

<table>
<thead>
<tr>
<th></th>
<th>$H$ (m)</th>
<th>$\theta$ (deg)</th>
<th>$T_{\text{reac}}$ (ms)</th>
<th>$T_{\text{prep}}$ (ms)</th>
<th>$T_{\text{step}}$ (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hsiao-Weckler et al. (2007)</td>
<td>1.63</td>
<td>12.5</td>
<td>75</td>
<td>150</td>
<td>100</td>
</tr>
<tr>
<td></td>
<td>1.63</td>
<td>17.5</td>
<td>75</td>
<td>150</td>
<td>125</td>
</tr>
<tr>
<td></td>
<td>1.63</td>
<td>21.6</td>
<td>75</td>
<td>150</td>
<td>150</td>
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<tr>
<td></td>
<td>1.63</td>
<td>27.5</td>
<td>75</td>
<td>150</td>
<td>225</td>
</tr>
<tr>
<td>Cyr et al. (2009)</td>
<td>1.73</td>
<td>30.7</td>
<td>75</td>
<td>150</td>
<td>175, 400, 275</td>
</tr>
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</table>
Table 3: The controller parameters related to cost function and constraints

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>No of samples, N</td>
<td>40</td>
</tr>
<tr>
<td>Sampling Time, T</td>
<td>25 ms</td>
</tr>
<tr>
<td>Horizon length, N×T</td>
<td>1 s</td>
</tr>
<tr>
<td>Simulation Time</td>
<td>2 s</td>
</tr>
<tr>
<td>Weight coefficient, $w_1$</td>
<td>$10^{-2}$ $s^2$</td>
</tr>
<tr>
<td>Weight coefficient, $w_2$</td>
<td>$10^{-2}$ $s^{-1}$</td>
</tr>
<tr>
<td>Foot length, $l_f$</td>
<td>c.f. Table 1</td>
</tr>
<tr>
<td>Max foot velocity, $v_{max}$</td>
<td>6 $m.s^{-1}$</td>
</tr>
</tbody>
</table>
Table 4: Predicted step lengths for different weightings of the cost function in the four balance recovery scenarios from Hsiao-Wecksler and Robinovitch (2007)

<table>
<thead>
<tr>
<th></th>
<th>0.32</th>
<th>0.50</th>
<th>0.73</th>
<th>1.19</th>
</tr>
</thead>
<tbody>
<tr>
<td>$w_1$</td>
<td>$10^{-2}$</td>
<td>$10^{-1}$</td>
<td>12.5</td>
<td>17.5</td>
</tr>
<tr>
<td>$w_2$</td>
<td>$10^{-2}$</td>
<td>$10^{-4}$</td>
<td>0.32</td>
<td>0.50</td>
</tr>
<tr>
<td>$w_3$</td>
<td>$10^{-4}$</td>
<td>$10^{-2}$</td>
<td>0.31</td>
<td>0.54</td>
</tr>
<tr>
<td>$w_4$</td>
<td>$10^{-2}$</td>
<td>$10^{-2}$</td>
<td>0.32</td>
<td>0.50</td>
</tr>
<tr>
<td>$w_5$</td>
<td>$10^{-1}$</td>
<td>$10^{-2}$</td>
<td>0.30</td>
<td>0.46</td>
</tr>
</tbody>
</table>
Figure 1: The two representations used of the human body. (a) Mechanical model: simple inverted pendulum + foot model, i.e. the CoM follows a circular arc. (b) Internal model: linearized inverted pendulum, i.e. the CoM travels at a constant height $h$. 
Figure 2: The feedback loop used to simulate the balance recovery.
Figure 3: Step lengths for single step recovery scenarios from Hsiao-Weckslar and Robinovitch (2007): experimental (white bars, averaged across subjects ± one standard deviation) versus simulated (black bars) results.
Figure 4: Stride length for multiple step recovery scenario from Cyr and Smeesters (2009). Experimental (white bars, averaged across subjects ± one standard deviation) versus simulated (black bars) results. The third step is not predicted by our model but was only observed for 2 out of 28 subjects.
Figure 5: Evolution of the *mechanical* model during the predicted recovery for the scenario of Cyr and Smeesters (2009) (snapshots every 200 ms).