Cooperative Control Design for Robot-Assisted Balance During Gait

Kooperative Regelung robotischer Gleichgewichtsunterstützung während des Gehens

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Summary Avoiding falls is a challenge for many persons in aging societies, and balance dysfunction is a major risk factor. Robotic solutions to assist human gait, however, focus on average kinematics, and less on instantaneous balance reactions. We propose a controller that only intervenes when needed, and that avoids stability issues when interacting with humans: Assistance is triggered only when balance is lost, and this action is purely feed-forward. Experiments show that subjects who start falling during gait can be uprighted by such feed-forward assistive forces. ▸▸▸ **Zusammenfassung** Sturzvermeidung ist eine ständige Herausforderung für viele Menschen in alternden Gesellschaften, und Gleichgewichtsprobleme stellen ein bedeutendes Risiko für einen Sturz dar. Heutige robotische Lösungen zur Unterstützung des menschlichen Gangs sind allerdings hauptsächlich auf das gemittelte Gangbild ausgerichtet und weniger auf instantane Gleichgewichtsreaktionen. Wir schlagen hier einen Regler vor, der nur dann interveniert, wenn nötig, und der keine Stabilitätsprobleme verursacht, wenn er mit dem Menschen interagiert: Stellgrößen werden erst dann generiert, wenn der Mensch das Gleichgewicht verliert, und diese Unterstützung fungiert als reine Vorsteuerung. Experimentelle Ergebnisse zeigen, dass menschliche Probanden, die während des Gehens zu fallen beginnen, durch solche unterstützenden, vorgesteuerten Kräfte in eine aufrechte Haltung gebracht werden können.

Keywords Balance, gait rehabilitation, exoskeletons ▸▸▸ **Schlagwörter** Gleichgewicht, Gangrehabilitation, Exoskelette

1 Introduction

Falling is an urgent challenge in our aging society, as falls are among the most frequent causes of hospitalization and death among the elderly [1]. Aside from muscle weakness, a key factor leading to falls is degraded balance control capability [2].

Humans rely on a fine interplay of strategies to maintain balance during standing and during locomotion [3; 4]: The "ankle strategy" moves the center of pressure (CoP), the point where the line of action of the net ground reaction force intersects with the ground [5]. The "hip strategy" moves the upper body in opposite direction with respect to the lower body, effectively changing the body's total angular momentum. Ankle and hip strategy are dominant during stance [3; 4]. The strategy that is predominantly used for balance control during locomotion is to adjust foot placement, if necessary inserting additional small steps [6; 7]. Also, the arms play a role in maintaining balance during gait (particularly in balance recovery [8]) by changing angular momentum.

Diverse technological solutions have been proposed to assist human balance: Training devices for walking [9; 10] or standing [11] can be connected to a base or an inertial frame, allowing almost arbitrary external forces and moments to assist the subject. Also portable solutions exist, mostly in form of exoskeletons, which can assist individual joints, for example to support ankle or hip strategy, or assist proper foot placement [12; 13]. Recently, we also proposed a balance-assisting backpack containing control moment gyroscopes [14].

To quickly identify situations where balance assistance is needed, important work has already been done in the area of fall detection, both in terms of theoretical modeling [6; 15] and sensor instrumentation [16]. Less is known on how balance-assisting control strategies should be designed such that they are cooperative and integrate well with human control actions.

Two major points need to be considered during control design for a balance-assisting device: First, a cooperative robotic support system should not override human control, because the human would adapt to the robotic support and increasingly rely on it [17]. To avoid such maladaptation, a device should only assist as needed [17; 18], providing just the support necessary to fulfill a task, which in this case is to recover balance. Second, whenever two controllers generate input signals to a system in a closed-loop manner and simultaneously, missing coordination between them can compromise stability. Therefore, the robotic controller should be designed such that it does not work against human control actions.

As a possible solution to fulfill these requirements, we suggest a strategy where the robotic device only generates *open-loop assistance*, which is *triggered at the instant when loss of balance is detected*. The feed-forward trajectory is calculated based on a model of the falling human, and it is designed such that it uprights the person given that model assumptions are true. We also show how such a control law can be implemented in a computationally efficient way when simplifying assumptions on the dynamics are made. Finally, we describe a simple experimental study on the Lokomat gait rehabilitation robot [19], to evaluate how humans interact with and perceive the assistance.

2 Feed-Forward Uprighting Control

2.1 Control Design

Two unsynchronized closed-loop controllers (human and robot) act on the same plant (the human body), which can lead to stability problems. To prevent this, we avoid closed-loop control, and instead provide only a feed-forward moment trajectory to the human body.

A model is needed to calculate the needed feed-forward trajectory. The equations of motion of a biped can be derived by summing moments about the center of mass [15], or about a fixed point [20; 21]. For a fixed point A on the ground (Fig. 1), the equations of motion are:

$$\dot{\boldsymbol{H}}_A = \sum \boldsymbol{M}_A = \boldsymbol{r}_s \times (m_h \boldsymbol{g}) + \boldsymbol{M}_{ext}\,, \qquad (1)$$

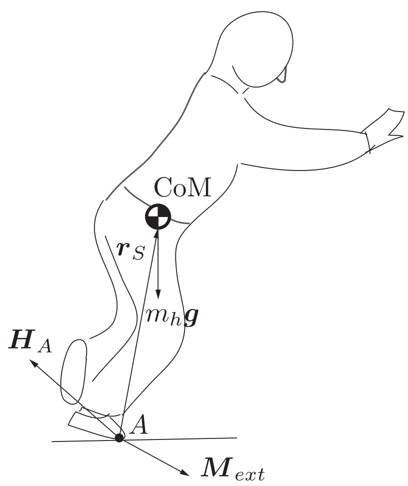

Figure 1 General model of a falling human: With respect to the fixed point A, the motion is characterized by the net angular momentum vector $\boldsymbol{H}_A$ and its rate of change $\dot{\boldsymbol{H}}_A$ across all body segments. While the gravity vector $m_h\boldsymbol{g}$ acting on the center of mass CoM promotes falling, a robotic device can create an uprighting moment $\boldsymbol{M}_{ext}$.

with $\boldsymbol{r}_s$ denoting the vector from A to the center of mass (CoM), m_h the mass of the body, and $\boldsymbol{g}$ the gravity vector. The angular momentum $\boldsymbol{H}_A$ is summed over all body segments, and its overall rate of change is $\dot{\boldsymbol{H}}_A$. A robotic device, for example a stationary training device, can exert an external moment $\boldsymbol{M}_{ext}(t)$ on the human. Thereby, one goal could be to bring the body to a stable posture with zero linear and angular velocity. Another goal could be just to slow down the fall, such that the patient has more time to regain control, for example to bring the swing leg to the front after a stumble.

With a model of the body's mechanics and of the sensorimotor control, measurement of the initial states at $t = 0$ (the instant the controller takes action), and a suitable parameterization of a moment profile, the system of differential equations (1) can be solved to find the assisting moment profile necessary to stabilize human posture.

In a simplified model, where the falling human is represented as an inverted pendulum and only small angles are considered (Fig. 2 left), the equations of motion can be drastically simplified to scalar form [22–24]:

$$m_h \ddot{x} = \omega_0^2 m_h x - F_{ext}\,, \quad \text{with } \omega_0 = \sqrt{\frac{g}{l}}\,, \qquad (2)$$

where l is the length of the pendulum, m_h is the mass of the human, x is the lateral displacement of the center of mass (CoM), and F_{ext} is an external force acting on the mass. It has been shown that despite their simplicity, inverted pendulum models are meaningful to analyze balance state [6; 15] and to derive control rules [22; 25] for human and bipedal robot balance [24; 26].

In particular, the "capture point" [22; 23] or "extrapolated center of mass" (XCoM) [6] are two equivalent concepts that have been derived from this model and

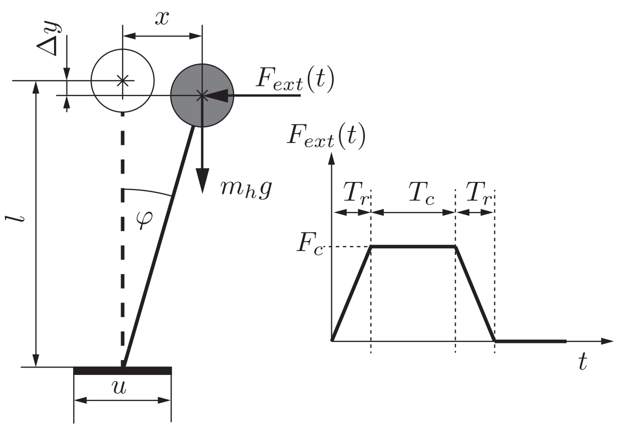

Figure 2 Control design: The human body is modeled as a linear inverted pendulum, i.e. a point mass m_h on a massless leg of length l and base of support of width u. For small angles φ, vertical displacement Δy can be neglected, and $x \approx l\varphi$. Based on this model, a feed-forward assisting force $F_{ext}(t)$ is found, to upright the human from given initial conditions.

proposed as measures to indicate the state of balance, with

$$\text{XCoM} := \dot{x}_0\omega_0^{-1} + x_0 \,. \tag{3}$$

Based on the linear inverted pendulum model, the location of the XCoM/capture point gives information about whether or not a person can still bring the body back up to static equilibrium in upright posture or not, given the initial states x_0 and $\dot{x}_0$. As long as the XCoM stays within the base of support (BoS), the person can return to the upright position by shifting the ZMP within the BoS (using ankle torque). If the XCoM leaves the BoS but remains within reach of a swing leg, the foot could be placed on that point to return the body to upright equilibrium within this next step. A main advantage is that the criterion relies only on kinematics and does not require measurement of ground reaction forces.

In case the subject can not stabilize posture anymore, an external force as in (2) will now be designed based on the same linear model. The idea is to apply a feed-forward profile, but it is unclear which shape is most intuitive for a human. In a first step, we chose a parameterized trapezoidal force profile, such that the force stays at a constant level F_c over a period of time T_c, and gently ramps up and down for a period T_r in the beginning and in the end (Fig. 2 right):

$$F_{ext}(t) = \begin{cases} F_c \dfrac{t}{T_r} & \text{if } 0 \le t < T_r \\ F_c & \text{if } T_r \le t < T_r + T_c \\ F_c \dfrac{2T_r + T_c - t}{T_r} & \text{if } T_r + T_c \le t < 2T_r + T_c \\ 0 & \text{else}\,. \end{cases} \tag{4}$$

Given initial conditions, a force F_c, and the ramp time T_r, a duration T_c of constant force application can be found that brings the human to desired final conditions, for example upright posture and zero velocity. This requires solving the differential equation (2) in piece-wise manner, leading to a two-point boundary value problem. It can be shown that a unique solution exists if the initial conditions x_0 and $\dot{x}_0$ satisfy

$$\text{XCoM} = \dot{x}_0\omega_0^{-1} + x_0 \neq 0 \tag{5}$$

(which means that the pendulum will not return to an upright position by itself), and if the constant force F_c is chosen such that it satisfies

$$F_{min} \le |F_c| \le F_{max}, \text{with} \tag{6}$$

$$F_{min} = \frac{T_r m_h \omega_0^3 |\dot{x}_0\omega_0^{-1} + x_0|}{1 - e^{-\omega_0 T_r}} \tag{7}$$

$$F_{max} = \frac{T_r m_h \omega_0^3 |\dot{x}_0\omega_0^{-1} + x_0|}{e^{-2\omega_0 T_r} - 2e^{-\omega_0 T_r} + 1}\,. \tag{8}$$

Then, the solution can be given in closed form:

$$T_c = \frac{1}{\omega_0} \ln\left(\frac{e^{-\omega_0 T_r} - e^{-2\omega_0 T_r}}{1 - e^{-\omega_0 T_r} - \frac{m_h T_r \omega_0^3}{F_c}(\dot{x}_0\omega_0^{-1} + x_0)}\right). \tag{9}$$

2.2 Experimental Setup

To experimentally evaluate the feed-forward uprighting controller, we conducted a small study with the Lokomat gait rehabilitation robot (Hocoma AG, Volketswil, CH). In the experiment, five non-impaired human subjects (3 m, 2 f, aged 26–60 y) walked on a treadmill, while the uprighting feed-forward controller assisted their balance.

The Lokomat is a stationary exoskeleton, originally designed to assist hip and knee flexion/extension during treadmill gait, while constraining leg and pelvis motion to the sagittal plane [19]. Vertical translation is not actuated, but the weight of the exoskeleton is passively compensated by a spring mechanism. In a modified setup at the University Hospital Balgrist, Zürich, the Lokomat has three additional degrees of freedom that allow the pelvis to move laterally during gait, and the legs to abduct and adduct [19; 27]. The rotation axes for hip flexion/extension and ab/adduction intersect approximately in the center of the human hip joint. In this experiment, dynamics of the robot that could have influenced ab- and abduction movements were passively compensated by springs, following the method in [28]. Lateral translation was actuated, so that the device was able to apply balance-assisting forces to the pelvis in lateral direction. A force sensor between the lateral actuator and the pelvis orthosis enabled force control with a RMS force tracking error of 13 N. The sagittal-plane actuators of the robot were in zero-force control mode, minimally interfering with the subject's movement [29].

In order to account for inertia of the robot in lateral direction, the controller had to be modified slightly: As the exoskeleton weight is supported by horizontal linear bearings, there is no additional gravitational moment related to lateral excursion. However, the exoskeleton did impose inertial forces on the subjects when they moved laterally, caused by residual apparent actuator inertia and exoskeleton mass. This additional inertia, but without additional weight, can be accounted for by changing the value ω_0 in (6)–(9) to the modified variable

$$\tilde{\omega}_0 = \sqrt{\frac{g m_h}{(m_h + m_r) l}}, \tag{10}$$

with m_h the human mass, and m_r the laterally moving mass of the robotic exoskeleton.

To create a challenging environment, where balance assistance would be needed during gait as well, subjects had to place their feet on a white line in the sagittal plane, which was drawn with chalk onto the treadmill, eliminating subjects' possibilities to use foot placement to control lateral balance by extending the BoS (Figs. 3 and 4). This created an environment similar to walking on a thin beam. Subjects were instructed not to hold on to the bars of the Lokomat environment, which would have provided additional stabilization. Each subject walked for 2 minutes, at a treadmill speed of 3 km/h.

The assistance triggered when the XCoM left the BoS, which had only a constant width of one foot (approximately 12 cm), due to the constrained foot placement. The preferred maximum value of the force profile was set to 100 N, but was adjusted if necessary to fulfill (6). A finite state machine ensured that the assistive force profile could not be triggered again before it was completely applied, and also that the system would pause for at least 0.2 seconds before the next application. Virtual and mechanical endstops would stop the subject's lateral excursion in case the assistance failed.

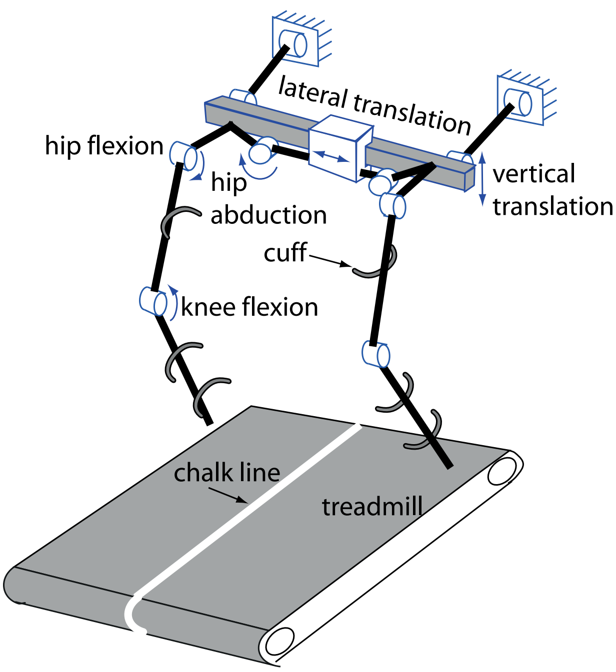

Figure 3 Schematic of the experimental setup: The modified Lokomat robot allows hip abduction and also lateral translation of the human pelvis. During the experiment, the subjects were instructed to place their feet exactly onto a chalk line.

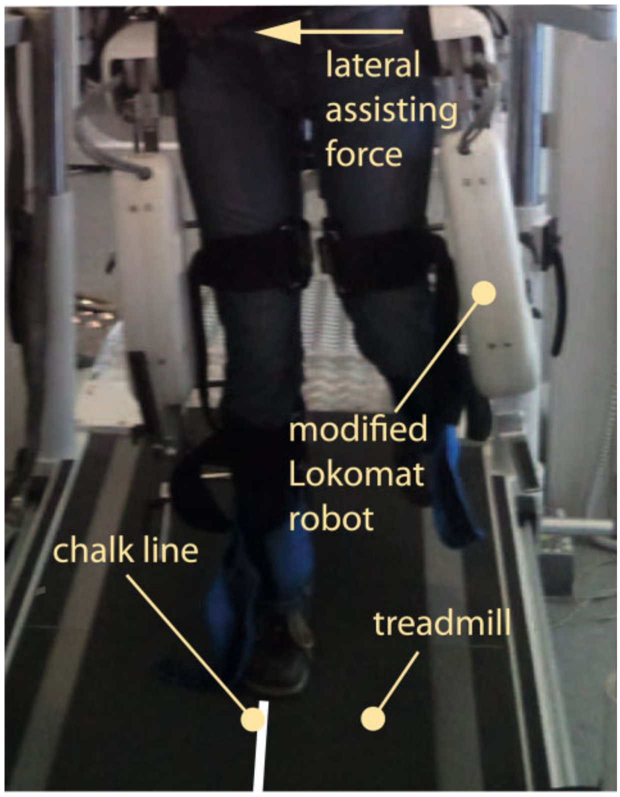

Figure 4 Photo of the experimental setup. All actuators were in zero-force control, except for the lateral pelvis translation, which provided balance assistance.

To discount transients and learning affects, only the last 30 seconds of the recorded 2 minutes were analyzed. Upon completion of each controller application, i.e. at $t = 2T_r + T_c$ in (4), we calculated the location of the XCoM with respect to the BoS. In case the XCoM was within the boundaries, the intervention was deemed successful, in case it had left the BoS again on the same side as before the intervention, the intervention was defined as too weak, and in case it had left on the opposite side, the intervention was defined as too strong. We then calculated the percentages of interventions for each subject that were judged successful, too weak, or too strong, and subsequently determined the mean and standard deviations across subjects.

3 Results

3.1 Simulation Results

In the phase plot (Fig. 5), both the XCoM criterion and the feed-forward controller action can be illustrated graphically: The allowed region for the XCoM is enclosed by the two lines $\dot{x}_0\omega_0^{-1} + x_0 > -u/2$ and $\dot{x}_0\omega_0^{-1} + x_0 < u/2$, with u being the width of the pendulum's BoS. Within this area, the inverted pendulum can still be uprighted

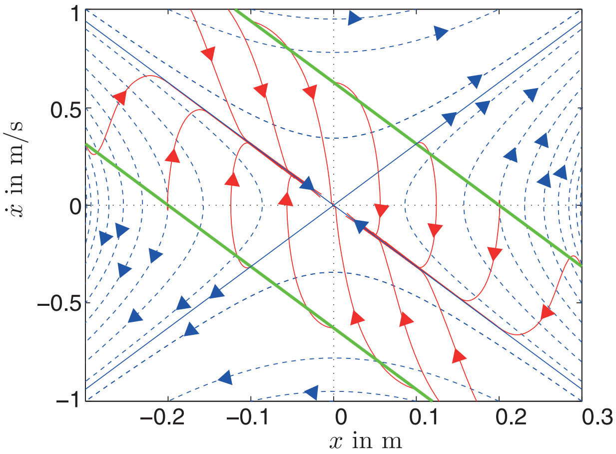

Figure 5 Phase plot representing controller action: Without control, the inverted pendulum behaves according to the dotted (blue) lines. The feed-forward controller acts when the system reaches the boundaries of stability, as defined by (3) and represented by the thick (green) lines for an exemplary base of support of width 0.4 m. The assisting controller pushes the system towards the center of the phase plot, which is the stable upright position, along the solid (red) lines.

by shifting the center of pressure within the BoS, so the controller remains inactive. However, once the system reaches the boundaries of this area, the assistive controller reacts and applies a feed-forward profile to the pendulum, uprighting it again.

3.2 Results of the Case Study

Across all five subjects, 95% of the controller interventions were evaluated as correct, with 11% standard deviation between subjects. None of the interventions were too strong, but 5% were too weak, with a standard deviation of 11%. The mean value of the XCoM after each intervention was 2.6 cm, thus smaller than a quarter of the width of the BoS, with a standard deviation of 1.3 cm between subjects. An exemplary time course of XCoM and force is shown in Fig. 6.

Two out of the five subjects found the controller too strong, and two others reported that it acted too early. In pilot controller tests preceding this study (with other subjects), two participants had also complained about discomfort and lower back pain. However, this did not occur during the study.

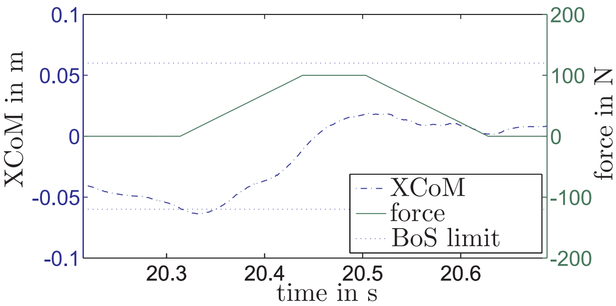

Figure 6 Example for a successful intervention during the study: The controller intervention brings the XCoM back into the BoS.

4 Discussion

In the experiments, the feed-forward controller provided effective balance assistance, bringing subjects back to a balanced posture. However, it might not be necessary to return the human to a fully vertical posture. Instead, it might be more comfortable to simply reduce the speed of falling, resulting in more time for the human to react. As elderly fallers have prolonged reaction times compared to non-fallers [30; 31], short-duration balance assistance may be already beneficial, bridging the gap of the first milliseconds before the patient is able to react.

Finally, the human model is overly simplistic. Future research will have to investigate more closely how humans react to external uprighting forces, for example in terms of reflexes.

5 Conclusion

We showed that feed-forward uprighting control can, in principle, help subjects restore balance during gait. The next step is to iteratively improve the controller, and to conduct experiments with a larger number of subjects. The outcomes can then be used to derive training protocols for gait rehabilitation using stationary robots, and also to develop wearable robotic systems that provide effective balance assistance in a patient's daily life.

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