BALANCE

Balance Augmentation in Locomotion, through Anticipative, Natural and Cooperative control of Exoskeletons

Deliverable 2.1

VPP and FPS during perturbed human standing and walking

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List of Acronyms

Acronym	Meaning
AP	Anterio-posterior
BSLIP	Bipedal SLIP
BW	Body weight
COM	Centre of mass
DOF	Degree(s) of freedom
DOW	Description of work
EMG	Electroymyogram
EXP	Experimental based research
FMCH	Force modulated compliant hip
FPS	Foot placement strategy / strategies
GRF	Ground reaction force
HAM	Hamstrings (muscle)
HAT	Head-arms-trunk
IP	Inverted pendulum (gait model)
LED	Light emitting diode
ML	Medio-lateral
MOD	Model based research
MVC	Maximum voluntary contraction
PP	Perturbation Platform (at TUDLL)
RF	Rectus femoris (muscle)
TUDLL	Technische Universität Darmstadt Lauflabor
SD	Standard deviation
SLIP	Spring-loaded inverted pendulum (gait model)
UTWEN	University of Twente
VBLA	Velocity based leg adjustment
VPP	Virtual pivot point
VPPC	Virtual pendulum postural control
WP	Work package
XCOM	Extrapolated centre of mass

1 Executive summary

This document describes various experimental and modelling studies conducted in order to investigate balance control mechanisms with respect to foot placement strategies (FPS) and the virtual pivot point (VPP). Experiments in both perturbed standing and walking were performed. In standing, external horizontal forces were applied in the sagittal plane. The resulting muscle activity to withstand the perturbations was analyzed and compared with a predictive model. Model predictions of bi-articular muscle activity were reflected in the EMG patterns, observed in healthy subjects. In walking, perturbations were applied in four directions in the horizontal plane, revealing that foot placement adjustment is not always necessary to counter the effects introduced by the perturbations.

In the analysis of human data the focus is on FPS and ground reaction force (GRF) magnitudes, while the VPP concept is mainly applied in model approaches. The VPP could be implemented in a walking model by regulating hip forces based on the forces along the legs. This allowed trunk stabilization with internal information only, i.e. without any information on absolute trunk orientation. Furthermore, stable walking patterns could be generated in a simple spring loaded walking model with asymmetric leg configurations, which lead to turning. Finally, a comprehensive literature review regarding research on perturbed standing and walking may help identify and design new experiments and modelling approaches. In that respect a review paper is being prepared in which a complete table of possible and accomplished studies on different kinds of perturbations in standing and walking will be presented.

This document gives concise descriptions of the various presented topics. References are made to the annex in which more detailed information can be found. Whereas the content of this document can be made publicly accessible, the annex should be treated **confidentially**. Some of its content consists of materials are scheduled for publication in the future or are currently in review.

2 Introduction

The aim of BALANCE Task 2.1 relating to this deliverable is to gain more insight in FPS and the VPP in balance recovery during standing and walking. In walking, foot placement is considered the most important strategy to maintain balance [MacKinnon et al. 1993], while in standing a change in base of support is also required if other strategies no longer suffice [McIlroy et al. 1996]. Consequently, foot placement is an important concept in regulating the GRFs to maintain and restore balance. Related to the GRFs is the VPP concept, which states that the ground reaction force vectors during unperturbed human walking intersect in a single point above the body's centre of mass. This behavior is observed in human walking, and was suggested to have stabilizing properties [Maus et al. 2010]. Derived from this observation, the VPP supplies a constraint on the direction of the GRF vector in modelling approaches. Here, the VPP was mostly applied in models, whereas FPS were mainly investigated by means of perturbation experiments in healthy human subjects.
Several experimental studies were proposed in T2.1 to investigate these concepts, such as standing and walking with horizontal perturbations, as well as walking on a rotating surface. All these studies rely heavily on a Stewart platform for applying perturbations. However, such a setup was deemed unsuitable given a Stewart platform's limited range of motion. A destabilizing acceleration must be shortly followed by a potentially stabilizing deceleration. This could influence the balance recovery responses and hamper data analysis. Therefore, new experimental setups were developed with which experiments have been - and are being performed. Due to the setup development time and obtaining new ethical approval some work has been moved to D2.2, such as the planned perturbations during standing by UTWEN. Furthermore, the experiment to investigate the effects of pitch and roll during walking was not performed. A rotating walking surface modifies two parameters at the same time; the (ankle) joint angles as well as the height of the stepping surface. Instead effort is being put in understanding the more fundamental concepts of human balance control.

3 Scientific progress

3.1 Perturbation matrix – A literature review (TUDLL)

To identify missing experiments and simulation models which may help understand balance mechanisms for the recovery from mechanical perturbations, a review paper is being prepared. The paper covers experiments and models in both standing and walking, although the focus is on the latter. The approach has the following goals, also see figure 1:

1) Identify existing and missing insights regarding experimental (EXP) and simulation model (MOD) based research.
    a. Insights from existing EXP → prove insights by simulation models?
    b. Insights from existing MOD → prove insights by experiments?
2) Identify the importance of tasks in daily activities
3) Derive the need for further experimental and modelling work based on 1) and 2)

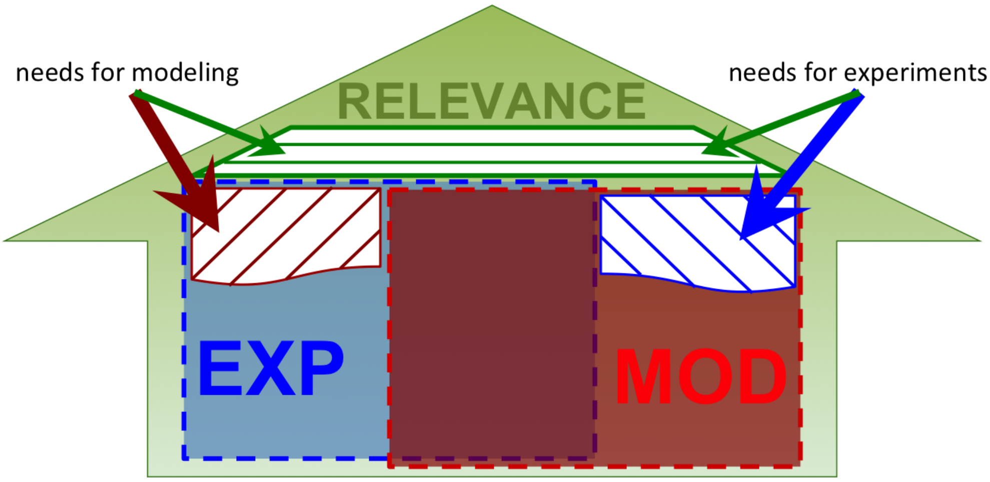


Figure 1: Graphical representation the perturbation matrix approach. Blue: publications found on experiments. Red: publications found on simulation models. Purple: intersection of the blue and red box, containing publications that complement each other. The relevance increase is shown by the green arrow and the shaded areas indicate need for experiments and models.

Table 1 shows various perturbation scenarios for standing and walking, and to which extend they are addressed in literature. A similar scheme was presented in the M18 report, suitable for showing existing and missing insights from literature. For daily activities, the following relevant research topics were identified in the literature:

Well researched

- EXP: Ground perturbation in walking
- EXP: External forces in standing

Not well researched

- EXP: Damped/elastic floor in walking
- EXP: External forces (push/pull) in walking.

Insights in balance mechanisms from experimental studies in perturbed walking on moving ground are for example:

- Step width is increased or decreased when the stance foot is laterally or medially perturbed respectively
- The foot might also be put down quickly with minimal change in step width
- Muscle groups show perturbation-specific responses

Furthermore, it becomes apparent that many works focus on parameter changes (e.g. step width) with the applied perturbations, rather than the relations between parameters for the different conditions.

Table 1. Perturbation matrix: the green (red) colour shows the topics which are (not) investigated.

Perturbations			Standing	*Walking*
Kind	**Feature**	**Properties**		
Ext. Forces	**Location**	***COM***	M – SU – H	E – CO – V
		Joint	E/M – SU – ML; E-SU-H	E – SU – ML
		Region	E-CO-H; E-SU-ML/V; E-CO/RA-H	
Ground Changes	**Steps/ Stairs**	***Up***	(Not applicable)	; E- CO/CH - R
		Down		E – SU; E- CO/CH - R
	Slopes	***Up***	E – SU/CH	E – CO
		Down		E – CO
		ML		
	Structure	***Elastic***		
		Damped		
		Stiction/ Friction		E – SU; M – SU; M – CO
	Moving	***H***	E - SU	E-SU
		ML		E – SU/CO
		V	E – SU/CH	E – SU
	Obstacle		(Not applicable)	E-SU; M – SU – H

Abbreviations:
Study type: E = human Experiments, M = simulation Models
Temporal structure: SU = sudden, CO = constant, CH = changing, RA = random
Direction of Perturbation: H = horizontal, ML = medio-lateral, V = vertical, AP = antero-posterior, R = random

3.2 [EXP+MOD] – Human leg morphology simplifies balancing (TUDLL)

Bi-articular muscles play an important role for balance control in the sagittal plane [Rode et al. in press]. These muscles provide the right combination of torques about hip and knee joints (bi-articular thigh muscles) as well as about ankle and knee joints (bi-articular shank muscle). This way perpendicular leg forces can be adjusted for the regulation of body position and orientation. Simulations show that results of conceptual models like the spring loaded inverted pendulum with trunk can be seamlessly transferred to models with segmented legs. Also, it is demonstrated that a natural distribution of motor tasks during standing (mono-articular knee extensors for load carrying, bi-articular thigh muscles in trunk orientation and synergetic action of all bi-articular muscles in body positioning) yields the so-called hip strategy in human standing. That is, the upright human can balance like a double pendulum with negative correlation of angular hip and ankle changes. The relevance of this concept is further supported by experiments in perturbed human standing, in which subjects were exposed to constant external pulling forces at the ankle and neck respectively, and were asked to keep posture.

Nine healthy subjects (age 24.3±1.8 years, mass 77.6±7.5 kg, height 1.82±0.07 m) participated in the experiment. The experiments were approved by the ethics committee of the Technische Universität Darmstadt and written informed consent was obtained from all subjects. Horizontal forces (10, 20, 30 N) were manually applied in the sagittal plane at the neck (anterior forces) and ankle (anterior and posterior forces), see figure 2. Subjects were instructed to keep their posture such that the muscles counteract the torques produced by the external force. During the experiment, subjects were asked to adjust the knee angle of the stance leg(s) (180°, 155°, 145°), which was checked with a sheet of cardboard. During ankle perturbations, subjects stood on one leg on the edge of an elevated force platform; the perturbation was applied at the straight free leg pointing downwards. Each experimental trial (one perturbation direction, one stance leg knee angle) consisted of six sub-trials applying randomized perturbation forces with each pulling force occurring twice. Each sub-trial comprised four phases: a resting phase (5s); a phase of pulling until the perturbation force was reached (4s); a hold phase (3s); a sudden release of the pulling force and subsequent balancing (3s). The order of experimental trials and sub-experiments was randomized for each subject. EMG signals of the hamstrings (HAM), rectus femoris (RF) and gastrocnemius muscles were collected (RF). Additionally, maximum voluntary contraction (MVC) experiments were performed.

When applying external forces to subjects, the predicted change in muscle activity reflected in the electromyographic (EMG) signal complies with measured EMG changes for bi-articular muscles, see figure 3. This is generally not the case for mono-articular muscles. Our theoretical and experimental results in combination with experiments on perturbed walking support the view that balance may be controlled universally and may not have to be learned separately for each motion condition. More details can be found in annex 1 [Rode et al. in press].

Figure 2. Experimental setup for perturbed human standing

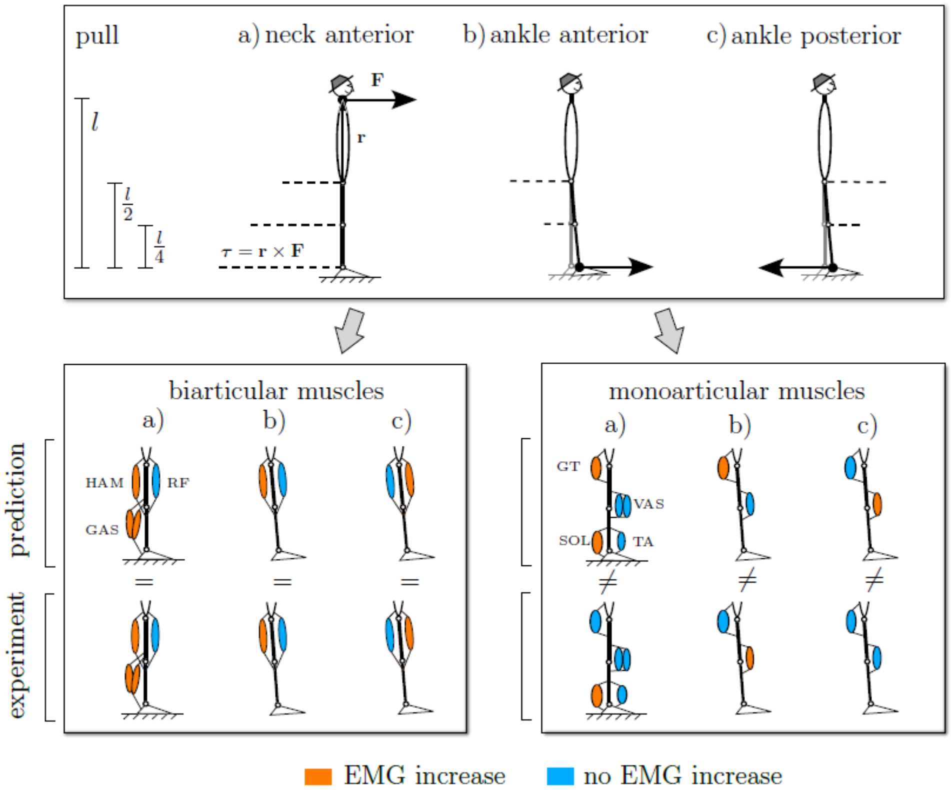


Figure 3. Experimental validation of the concept. Top: A static external force F results in torques τ (magnitude represented by length of dashed lines) about the leg joints to be counteracted by muscles to maintain posture at three different knee angles (180° shown, 155°, 140° not shown). Bottom: Orange muscles show significant EMG increases for the 3 configurations shown in the top row (a,b,c). In blue muscles no EMG increase was found. In contrast to mono-articular muscles, predicted EMG change of bi-articular muscles complies with measured EMG change. Across conditions, in seven out of ten cases bi-articular muscles revealed a significant positive linear relation between EMG increase and pulling force as predicted. For mono-articular muscles this was only true in two out of eight cases.

3.3 [EXP] – Swing leg role in balance (TUDLL)

Models assuming massless legs (e.g. based on SLIP or IP templates) have been widely used to interpret human gait. Although these models can describe basic gait features like CoM trajectories or ground reaction force (GRF) patterns, it is rather challenging to investigate swing leg control strategies with these massless leg models. For example, segment accelerations of swing and stance leg could also affect the GRF pattern, which is attributed to stance leg function by massless leg models. Investigating each body segment's contribution to GRF can help understand the observed leg function. In addition, massless leg models cannot describe human locomotion in the situation when human subjects are under perturbations which require keeping balance by accelerating limbs (e.g. the swing leg) and consequently creating a torque on the trunk. That is, accelerating limb mass can potentially help human subjects stabilize the trunk.

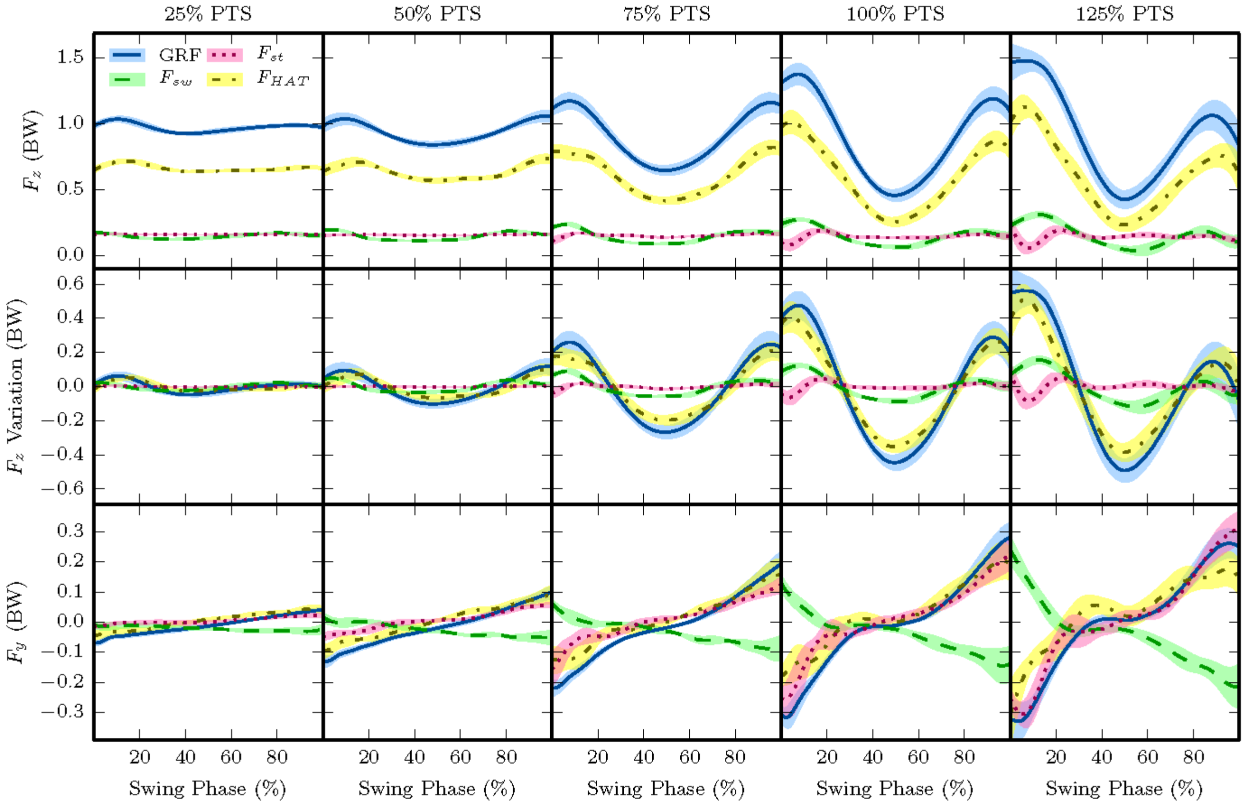


Figure 4. GRF components during human walking at different speeds (25-125% PTS). PTS denotes the preferred transition speed from walking to running. 75% PTS is about the normal walking speed. F_z and F_y denote the force in vertical and walking direction. GRF is the ground reaction force and F_{sw} , F_{st} and F_{HAT} denote the force contributions of swing leg, stance leg and HAT (head, arms and trunk), respectively. All forces are normalized to body weight (BW). Bands on each waveform indicate the standard derivation (±SD).

Therefore, in order to understand the effect of leg movement on the dynamic behavior of human walking, leg dynamics during single support phase were investigated by analysing the contributions of the leg and upper body accelerations to the total GRF. The results in figure 4 show that:

- The vertical force created by the upper body is dominant in all speeds. Furthermore, the swing leg also contributes to the characteristic M-shape force pattern in walking, in contrast to stance leg. For preferred walking speed (around 1.5m/s), swing leg dynamics explain about 25% of the typical M-shaped GRF variation.

- The horizontal forces created by the stance leg mass accelerations and HAT are similar in both magnitude and shape. In contrast, swing leg mass accelerations result in force contributions with similar shape and magnitude but opposite direction.

These results will help refine existing walking models by adding a trunk, rigid stance leg and spring like operating swing leg. This model will help to better describe human walking and possible reactions to maintain balance after perturbations. More details are described in annex 2 [Zhao et al. in press].

3.4 [EXP] – Foot placement adjustment in perturbed walking (UTWEN)

Various concepts such as the extrapolated center of mass (XCOM) [Hof, 2008] and velocity based leg adjustment (VBLA) [Sharbafi et al. 2014] suggest foot placement strategies related to the the velocity of the center of mass (COM). How the COM position and velocity relate to the foot placement location and swing time has not been thoroughly investigated for perturbed human walking. In this study, relations between the COM position and velocity, the foot placement location and the swing time were investigated for medio-lateral (ML) and anterio-posterior (AP) perturbed treadmill walking.

Ten healthy volunteers participated in the study. Subjects walked on a dual belt instrumented treadmill and randomly received horizontal perturbations at the pelvis, applied at right toe-off. Perturbation force was equal to 4, 8, 12 and 16% of the subject's body weight. Motion capture data, ground reaction forces and leg muscle EMG signals were collected. Instances of the first right heel strike after perturbation onset were identified, and the location of the feet relative to the COM were investigated, see figure 5 for a top-down view. Compared to unperturbed walking, most perturbations lead to changes of the COM position and velocity relative to the stance foot. While for ML perturbations the location of the new placed foot varied with perturbation magnitude, this was not the case for AP perturbations. However, the ground reaction forces delivered in the subsequent double support phase did change with perturbation magnitude. Hence, direct foot placement adjustment following a perturbation is not always required for recovery. For more detail, please refer to annex 3 [Vlutters et al].

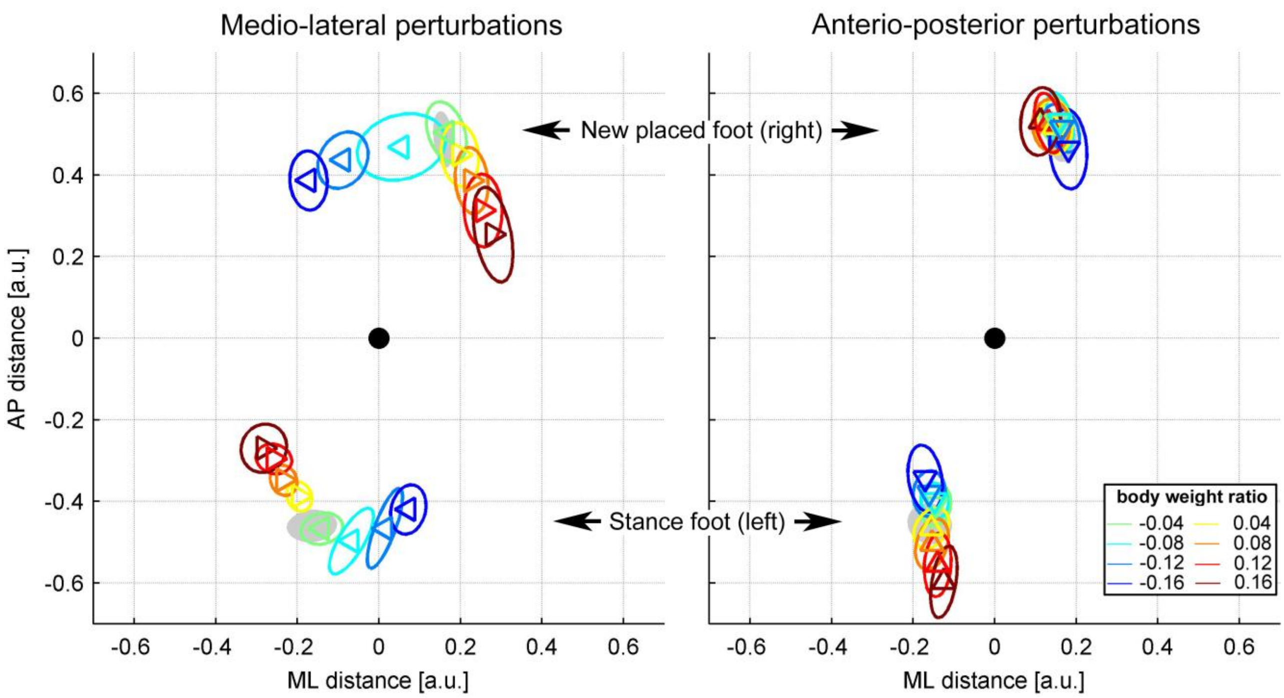
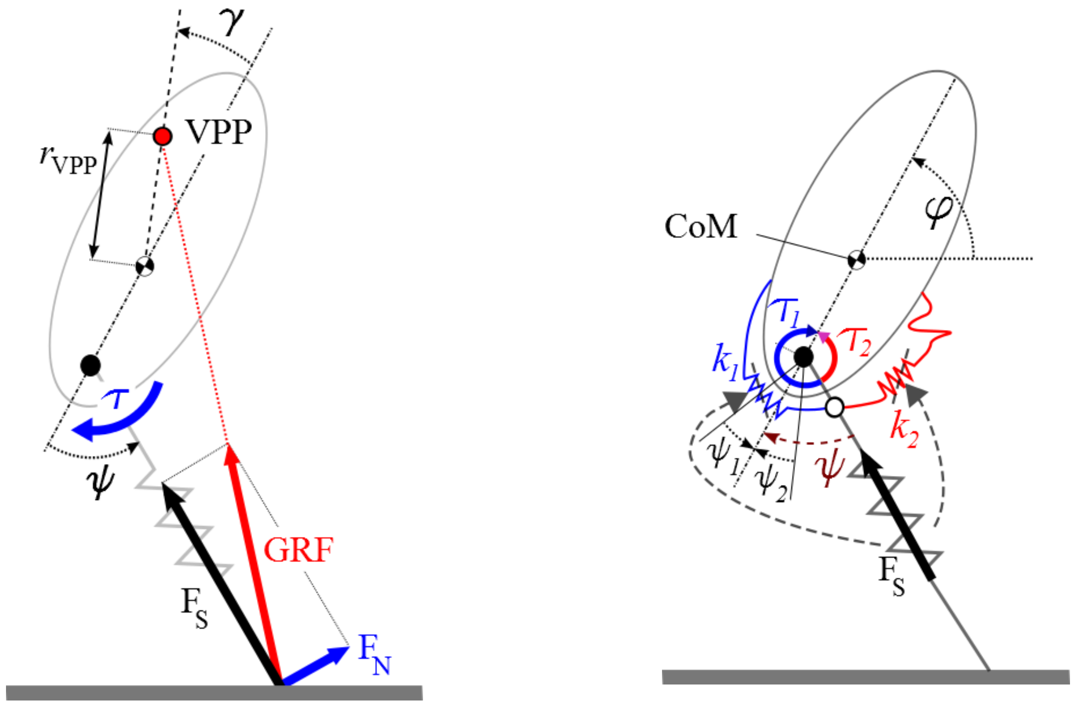


Figure 5. Top down view of the subject average location of the COM of each foot relative to the total COM (at 0,0) at right heel strike following the perturbations at right toe-off, for medio-lateral (left) and anterio-posterior (right) perturbations. Ellipses indicate the subject standard deviation. The direction of the triangles indicates the perturbation direction. The different colors relate to the different perturbation magnitudes. Gray areas indicate values for unperturbed walking. Walking speed was 2.25 km/h scaled to leg length. All quantities are dimensionless.

3.5 [MOD] – Force modulated compliant hip (FMCH) model (TUDLL)

Balancing the upper body as one of the main features in human locomotion can be achieved by actuation of compliant hip joints [Sharbafi et al. 2014]. The presence of a virtual pivot point (VPP) is characteristic to human walking with upright trunk posture. Based on the VPP concept, stable gaits can be found in a conceptual SLIP-based model extended with a rigid trunk, called TSLIP. Virtual pendulum based postural control (VPPC) generates hip torque such that the GRF remains directed toward the VPP, a fixed point on the upper body above the centre of mass (fig. 6a).
In a new model (fig. 6b), the VPP can be implemented through a leg force modulated compliant hip (FMCH). This implies that VPP based upright trunk posture could be achieved through sensory feedback from leg extensor force to activation of hip muscles. With this mechanism, the VPP concept could be translated into a neuro-mechanical system by modulating the direction of leg force based on elastic hip joint function. Please refer to annex 4, [Sharbafi et al. 2014] for more detail.



(a) **(b)**

Figure 6. TSLIP (Trunk+SLIP) model and two control approaches: **(a)** Virtual pendulum-based postural control (VPPC) and **(b)** leg force modulated compliant hip (FMCH).

3.6 [MOD] – 3D walking simulation studies (TUDLL)

Blindfolded or disoriented people have the tendency to walk in circles rather than in a straight line. Here a minimalistic walking model (fig. 7a) was used to examine this phenomenon. The bipedal spring-loaded inverted pendulum (BSLIP) exhibits asymptotically stable gaits with CoM dynamics and ground reaction forces similar to human walking in the sagittal plane. This model was extended into three dimensions, and show that stable walking patterns persist if the leg is aligned with respect to the body (i.e. the CoM velocity) instead of a world reference frame. Furthermore, it was demonstrated that asymmetric leg configurations, which are common in humans, will typically

lead to walking in circles (Fig. 7b). The diameter of these circles depends strongly on parameter configuration (e.g., difference in leg length), but is in line with empirical data from human walkers. Simulation results suggest that walking radius and especially direction of rotation are highly dependent on leg configuration and walking velocity.
Analysis of the model's walking stability, measured by the magnitude of the largest eigenvalue of the return map, is not only preserved but can be greatly enhanced in asymmetric configurations. This is in line with previous results from [Merker et al. 2011], where similar results were found for planar situations. Parameter deviations were analysed, and the structural stability of the model (up to energy) was established. More details can be found in annex 5, [Maus et al. 2014], which is now published in Proc. Royal Society Interface.

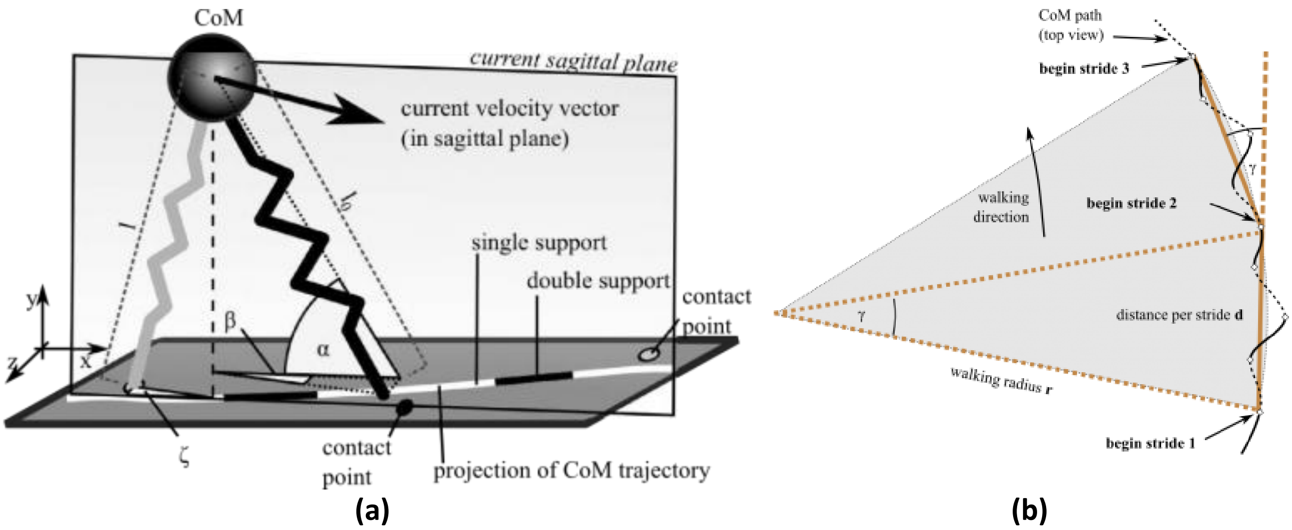


Figure 7. (a) Schematic drawing of the three-dimensional BSLIP model. During swing, the leg is oriented with respect to the current sagittal plane spanned by the velocity and gravity vectors. α and β denote the polar angle and azimuthal angle, respectively. ζ denotes the azimuthal angle between sagittal plane and stance leg. l_0 and l denote the legs' rest length and current length, respectively. **(b)** Sketch of a top view on the horizontal plane. The walking radius is the radius of the circle which connects all CoM positions at the beginning of a stride. Dashed and solid parts of the CoM path correspond to single and double support phases, respectively.

4 Overall discussion

To investigate balance control in perturbed human standing and walking, several key steps are being performed:

1- Observing human behavior in experimental studies
2- Model development to mimic human behavior
3- Formulating control strategies

After identifying the effect of different perturbations on human balance through experiments and/or simulation studies, a control strategy can be formulated based on the obtained results. Experimental and model work can complement each other and assert a formulated control strategy.

The presented work (sections 3.4, 3.5, 3.6) suggest potential for velocity based foot placement control, for both normal flat ground walking and perturbation recovery during walking. Models can walk stably using foot placement based on VBLA (3.5, 3.6), and experimental results (3.4) suggest a strong linear relation between ML foot placement and the ML COM velocity. However, the experimental results also showed that this relation does not hold for all perturbations, as foot placement remained largely unchanged for AP disturbances. Other factors such as the task which the subject is trying to execute (e.g. walking direction, initiation or termination of movement) likely play an important role as well. An extended control strategy for foot placement should therefore be formulated, which in turn has to be validated by new experimental and simulation studies. The performed literature review will help navigate this future research.

The VPP concept has successfully been applied in model simulations (sections 3.2, 3.5), where it is used as a control objective for upright stabilization in both standing and walking. It provides a constraint within which models can show human-like behavior. However, the implications of a VPP computed from human data remains uncertain, as guidelines for the quality of the computed VPP need yet to be developed. It is currently uncertain which deviations from the point can be tolerated, and which require specific control actions (e.g. in perturbation recovery). Further analysis of decomposed GRFs might yield more insight in the occurrence of this concept.

The presented results provide a first basis for formulating foot placement control and body stabilization strategies in human balance control. For example, following the experimental results (section 3.4), in case of excessive forward walking velocity the exoskeleton might increase leading leg stiffness and deliver breaking forces in the double support phase. Hip torques might be adjusted accordingly using FMCH (section 3.5) and keep the trunk upright. In other scenario's such as standing, the bi-articular muscle model (section 3.2) might help in stabilization by the prediction of exoskeleton joint torques. Finally, implementation of the control strategies into the exoskeleton leads to a fourth step:

4- Observing combined human and exoskeleton behavior in experimental studies

This will be the final project step, using the formulated evaluation scenario's. As both human and exoskeleton will act and react to each other, the performance of either will depend on the other. Less adaptation from the human side might be required if the exoskeleton "understands" how its user is trying to maintain balance. This stresses the importance of a thorough understanding of the human control strategies, to be provided by WP2.

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