Benchmarking gait models for simulating the influence of body weight unloading on human gait

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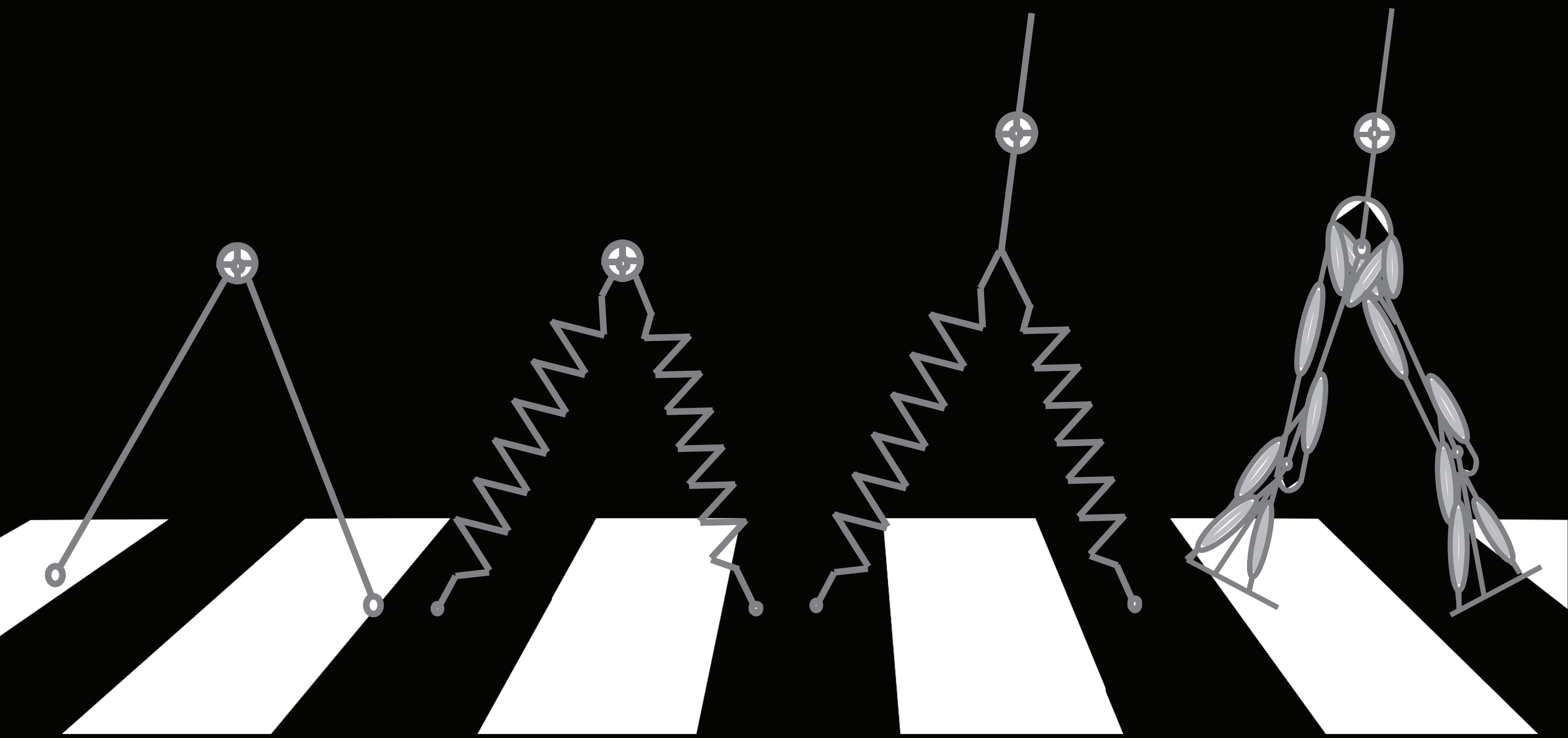

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Challenge the future

BENCHMARKING GAIT MODELS FOR SIMULATING THE INFLUENCE OF BODY WEIGHT UNLOADING ON HUMAN GAIT

by

Salil Apte

MSc. Thesis report
in partial fulfillment of the requirements for the degree of

Master of Science
in Mechanical Engineering

at the Delft University of Technology,
to be defended on Tuesday November 27, 2017.

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This thesis is confidential and cannot be made public until December 31, 2019.

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PREFACE

This thesis report is divided into two components, a research manuscript which is the main document and an appendix which provides the background information for this manuscript. The manuscript is intended for submission to the IEEE Transactions on Neural Systems and Rehabilitation Engineering journal and hence has three contributing authors. The author contributions are as follows: SA conducted the research, analyzed and interpreted the data and was the main contributor to the manuscript. MP and HV guided the research and scrutinized the results. In addition, MP and HV supervised the writing process for the manuscript.

Salil Apte
Delft, November 2017

Benchmarking gait models for simulating the influence of body weight unloading on human gait

Salil Apte, Michiel Plooij and Heike Vallery, *Member, IEEE*

***Abstract*—Modulated body weight support (BWS) systems have shown promise in improving the task specificity of BWS training. This promise, however, is based on pilot studies and limited number of modulation strategies for the unloading force. To evaluate the effectiveness of these strategies and select the most favourable strategies, simulation of the influence of modulated unloading force on gait models can be useful. However, this method is not commonly utilized. Since the reliability of the simulation results depends on the gait model, bench-marking of gait models based on their response to BWU is an important stepping stone in the design of modulated BWS systems. This work bench-marked the Simplest walking model, the Spring-loaded inverted pendulum model and the Muscle-reflex gait model based on their suitability for simulation of human walking under the influence of BWS. Three realizations of BWS, based on Constant force, Counterweight and Tuned spring approaches, were designed and the response of the models to BWU was compared with existing human experimental data. Out of the three models, the results suggest the SLIP model to be the most suitable for BWS simulations. This study may provide a springboard for the development of novel BWS strategies and help enhance the task specificity of BWS training.**

Keywords*—Body weight support, Gait models, Gait characteristics, Simulation, Rehabilitation

I. INTRODUCTION

Body weight supported training (BWST) is a viable gait rehabilitation technique for individuals suffering from neurological impairment due to diseases like stroke, spinal cord injury, Parkinson's disease, etc. During BWST, a certain amount of the user's body weight is supported by a suspension system typically through a harness worn by the patient [1]. After undergoing BWST, individuals have shown amelioration of balance, motor function and overall capacity of locomotion [2–8]. In addition to these benefits, BWST can lead to improved psychological well-being, enhanced muscle mass and better cardiovascular health [9]. Body weight support (BWS) systems allow clinicians to monitor gait rehabilitation, without the need of providing complete physical assistance [10].

A BWS system is typically composed of an apparatus which provides an unloading force to the user walking overground or on a treadmill [15, 16]. BWST as a clinical intervention relies on neuroplasticity of the brain [17], which makes it important that the gait which is trained under BWS is similar to the normal gait of the user. Though still debated [18], task specificity seems to be important to rehabilitation [19] and BWST tries to make use of such a task-specific approach. Apart from the walking environment, the nature and the magnitude of the BWU force play a major role in determining the gait of the user.

Unloading force can be broadly classified into two categories – (1) *unmodulated*: where the goal is to provide a constant magnitude of unloading force and (2) *modulated*: where the unloading force is modulated according to the specific gait parameters [20]. While the majority of BWS systems belong to the first category, the latter category has recently seen some novel and promising BWS systems [20]. Some examples include a BWS system which controls the unloading force based on gait cycle phases [21], another where the centre of pressure trajectory governs the unloading force [22] and also a system that aims to dynamically compensate the inertial forces of the user's body [23]. Modulation of unloading force can enable appropriate ground contact and limb motion while allowing gait spatio-temporal parameters like walking speed, cadence and stride length to be comparable to the values during unsupported walking [20]. Besides modulated unloading force, application of vertical and forward forces together has recently shown good results [24]. These results are promising and take a step further towards task-specific BWST, albeit with evidence limited to pilot studies. Furthermore, the design of modulation strategies and forward forces is a largely heuristic technique and doubts exist over its generalizability.

New BWS designs have been proposed recently [25] and more will be conceived in future that allow for better modulation of the force vector, also in other directions than the vertical. One way to test the efficacy of new BWS system designs and modulation strategies is by simulating their influence on locomotion of existing gait models. This can improve the efficiency of the design process by speeding up the iteration steps. This enables pruning of possible designs and modulation strategies to select the only the promising ones, which can later be translated to prototypical BWS systems and tested further through human experiments. Examples of such a approach are the studies by Glauser et al., Ma et al. and Lu et al. [26–28]. These examples, however, show that there is a wide range of gait models currently being used for such a simulation and they range from the simplest (mass-spring-damper system) to the most complex musculoskeletal models.

Considering the above background, the main goal of this research is to investigate the suitability of gait models for BWS simulation through a comparison with the experimentally-obtained gait features. Simulations are conducted in the sagittal plane. Section II details the three distinct BWS strategies considered in this research: (1) *Constant force* (CF): which emulates a constant vertical unloading force (2) *Counterweight* (CW): A vertically moving counterweight is used to provide

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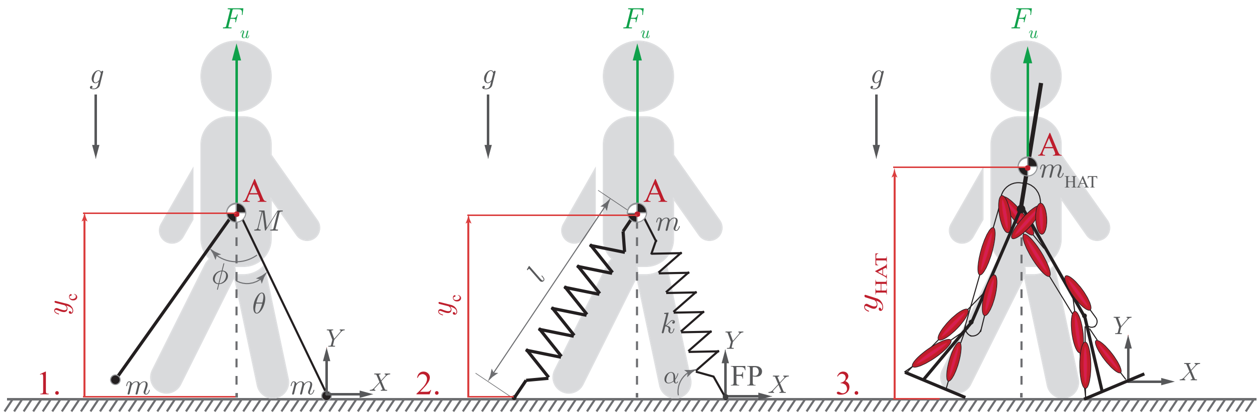

Fig. 1. The three gait models considered in this paper: (1) Simplest Walking (SW) model [11] where M is the mass of the body, m of each foot and m is assumed to be negligible as compared to M, θ is the stance leg angle w.r.t to vertical, y_c is the vertical position of the centre of mass and ϕ is the swing leg angle w.r.t to the stance leg. Details of the actuation principles from [12] are not shown here. (2) Spring-loaded inverted pendulum (SLIP) model [13] where m is the mass of the body, l is the original leg length, α is the angle-of-attack, y_c is the vertical position of the centre of mass, k is the stiffness of the leg spring and FP is the foot point of the stance spring. (3) Muscle-reflex (MR) model [14] where point **A** represents the centre of mass of the upper body, y_{HAT} is the vertical position of this centre of mass and m_{HAT} is the mass of the upper body. For all three models, the vertical unloading force F_u is applied at point **A**.

the unloading force and (3) *Tuned spring* (TS): an elastic element (spring) with specifically tuned stiffness generates the unloading force. The TS strategy is a recent development [29] and has not been commonly used, if at all.

Three prominent biomechanical gait models from literature are simulated with the above-mentioned gait strategies and trends for gait parameters are documented. The three gait models (Figure 1), in the increasing order of complexity are: (1) *Simplest Walking (SW) model* , (2) *Spring Loaded Inverted Pendulum (SLIP) model* and (3) *Muscle-reflex (MR) model* [11, 13, 14]. Hereafter, these models are referred using the respective abbreviations. The SW model is actuated on the basis of the principles suggested in [12] and the foot mass is assumed to be negligible as compared to the body mass. The equations of motion for these models, as used in the simulations, are presented in Section III. Gait parameter trends produced by the simulations are compared with the human data trends obtained from [20]. These results are shown in Section IV. Finally, the implications of our results and the limitations of the simulations are discussed in Section V.

II. Methods

A. Selection of BWS strategies

This section describes the three BWS strategies (Figure 2) used for simulations, CF, CW and TS. The main purpose of providing constant unloading force is to partially reduce gravity. The notion that constant force is the best solution for partial BWS has been dominating the field of BWS systems [20], and led to complex mechanical designs such as the Lokolift [15], the Zero-G [30], etc. These devices use active control in order to render a constant force. This is different from actual simulated gravity because the load is applied only at the upper body of the human (distributed via the harness), and not in a distributed way on each single body segment [31]. Accurate investigations for swing phase generally require setups like a parabolic space flight [32, 33], which are inconvenient to reproduce. The CW and TS strategies are commonly used in existing BWS systems to produce BWU force [15]. However, the spring stiffness for the TS system is selected on the basis of the hypothesis suggested in [29]. This way, the simulations also enable a first comparison of the TS approach to the other two and present a way of testing this hypothesis.

The simulations are based on four main assumptions (Figure 5) – (1) the counter-weight and the free end of the spring only move in the vertical (Y) direction, (2) pulleys **I** and **II**, the ropes and the spring in Figure 2 are massless, (3) the BWS system is frictionless and there is no energy dissipated in the system and (4) the pulley **I** follows the attachment point **A** along the horizontal (X) direction and thus it is always perfectly overhead of the attachment point. In other words, the X position of the pulley **I** and attachment point **A** is always the same. Thus, the BWS system does not apply any horizontal forces on the gait model nor does it add to the inertia of the model in horizontal direction. While the horizontal inertia of the BWS system [15] can be important for determining the user's gait, we chose to focus solely on the influence of the vertical unloading force on the gait. Considering the % BWU supplied to be β, $u = \beta/100$, total mass of the body to be M and the acceleration of gravity to be g, the equations describing three BWS strategies are presented below.

Constant force: The Constant force BWS strategy consists of applying constant vertical force ($F_u = uMg$) on the body without supplying any additional inertial forces. It can be considered as an ideal case of unmodulated BWU [20]. Since the SW and SLIP models do not have distributed mass, the CF BWS strategy also emulates the effects of reduced gravity for these models [31]. However, it is not the case for the MR model [14] due to the presence of limb mass. The CF strategy entails application of unloading force at the center of mass (COM) of usually the upper body while the reduced gravity affects all body segments.

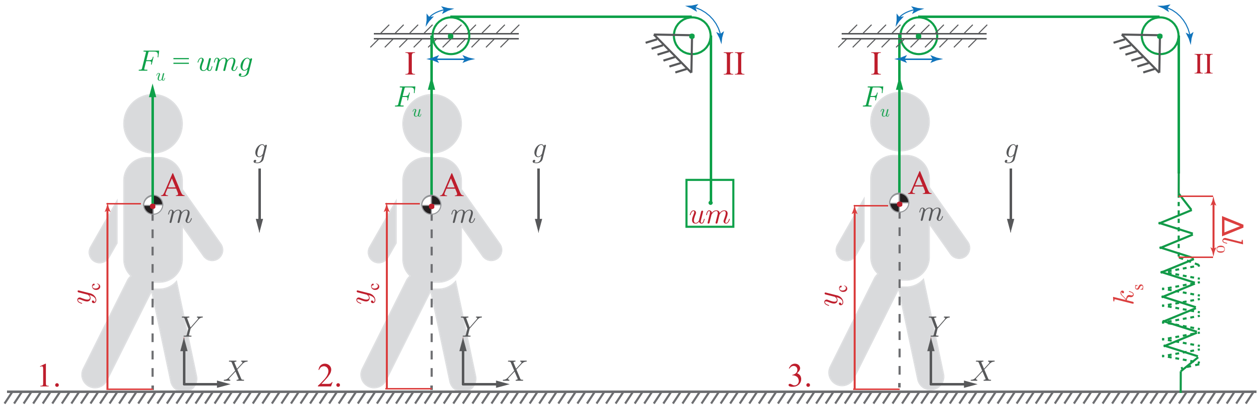

Fig. 2. Three BWS strategies: (1) Constant force (2) Counterweight system (3) Tuned spring system. Centre of the pulley **I** is assumed to move horizontally such that force F_u is directly vertically upwards from point **A**. Centre of pulley **II** is fixed and pulley **II** can only rotate. Both pulleys are massless and there is no energy dissipation in the system. y_{c} is the vertical position of the centre of mass, u is the amount of body weight unloaded as a proportion of the actual body weight, k_{s} is the stiffness of the spring and Δl_0 is its initial elongation.

Counterweight system: The Counterweight (CW) BWS strategy is based on the use of a counterweight of mass m ($m = uM$) to provide β % of BWU. This strategy leads to a constant unloading force ($F_u = mg$) where g is the acceleration due to gravity. However, the counterweight moves vertically as it follows the vertical motion of the attachment point **A**. Due to this motion, an additional inertial force ($m\ddot{y}_{\mathrm{c}}$) is generated, which disturbs the intended constant unloading force. Thus, instead of a constant unloading force, the force acting on the body is $F_u = mg - m\ddot{y}_{\mathrm{c}}$, where $\ddot{y}_{\mathrm{c}}$ is the vertical acceleration of the attachment point **A** in downward direction.

Tuned spring system: An elastic element, which can be considered massless as compared to a counterweight, can provide unloading force and reduce the problem of inertial forces caused by the movement of the counterweight. The motion of the attachment point, though, affects the deflection of this elastic element (spring), thus causing variations in unloading force. The Tuned spring BWS system is based on the concept of using a spring (elastic element) to provide an unloading force. While the unloading force compensates for the weight of the user, the inertia of the body still affects the dynamics of the gait. If the unloading force can be tuned to compensate for both the gravitational and inertial forces, gait dynamics will be less modified, thus improving the task specificity of the BWST. According to the hypothesis presented in [29], the stiffness of the spring used for providing the unloading force can be tuned to compensate for inertial forces of the unloaded mass. This works for a periodic (ideally harmonic) movement of the centre of the mass of the body. The stiffness of this spring in this case is given by:

$$k_{\mathrm{s}} = u\omega^2 M \tag{1}$$

where M is total mass, $\omega = 2\pi c$ and c is the cadence of the walking model at 0% BWU. The initial deflection of the spring (Δl_0) to produce unloading force is:

$$\Delta l_0 = \frac{uMg}{k_{\mathrm{s}}} = \frac{g}{\omega^2} \tag{2}$$

The unloading force provided by the TS BWS strategy is:

$$F_{\mathrm{ts}} = k_{\mathrm{s}}\Delta y = k_{\mathrm{s}}(y_{\mathrm{c}0} - y_{\mathrm{c}} + \Delta l_0) \tag{3}$$

where y_{c} is the vertical position of point A at time t and $y_{\mathrm{c}0}$ is its position at the start of the simulation ($t = 0$).

B. Selection of gait models

The scope of this research is limited to 2D gait models since all the gait characteristics of interest i.e. those studied in [20], can be investigated using 2D models. Initially, five prominent gait models were considered for this research: (1) Linear inverted pendulum model (LIPM) [34], (2) Simplest walking (SW) model [11], (3) Spring-loaded inverted pendulum (SLIP) model [13], (4) Virtual pivot point (VPP) model [35] and (5) Muscle-reflex (MR) gait model [14]. The LIPM model, however, considers the centre of mass (COM) of the body to move in a straight horizontal line and thus the vertical movement of the COM needed to study the counterweight and tuned spring BWS strategies is absent. As a result, this model was not present in the final selection of gait models. The original implementation of the VPP model could not be reproduced in a robust way and hence an alternative implementation [36] was tested. However, this implementation was based on a 'Capture Point' controller [37] instead of the 'Constant angle-of-attack' controller used in the original implementation. Taking this into account, the VPP model was not selected for the benchmarking study. The results for the 'Capture Point' based VPP model are available in the Appendix.

C. Simulation procedure

Each gait model was simulated with all three BWS strategies using Matlab and Simscape (MR model). All models were obtained online [38–40] and were modified according to the equations presented later in this section. Each modified model was simulated with BWU ranging from 0% to 100%, in 5% increments. The unloading force was applied at the center of mass of the body ($\mathrm{COM_{body}}$) for all gait models and the COM

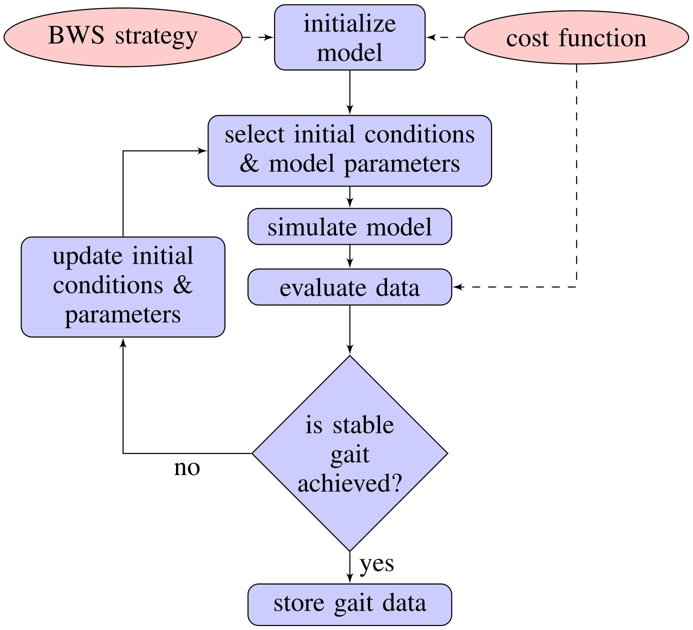

Fig. 3. Flowchart of the simulation process for SW and SLIP models

of the upper body for the MR model. At each level, the model parameters and initial conditions were optimized (Figure 3) to achieve a stable gait cycle [41] and consequently the longest walking distance. The highest percentage of BWU for which the model was able to achieve a stable gait was noted as the 'Maximum feasible BWU' ($\beta_{\max}$) for each strategy.

In case of the MR model, Simscape blocks were created to emulate the constant force (CF), counter-weight (CW) and tuned spring (TS) BWS strategies (see Appendix for details). A similar simulation process was followed for the MR model. The optimization procedure was skipped for the MR model due to the large number of model parameters and initial conditions.

For the MR model, the selection of the location where the unloading force acts is an important decision. Since the limbs in this model are assumed to have mass, the center of mass of the body ($\mathrm{COM_{body}}$) is different from the center of the mass of the upper body which includes the head, arms and trunk ($\mathrm{COM_{HAT}}$) and excludes the legs. The distance of the $\mathrm{COM_{body}}$ from the hip joint (d), along the length of the upper body, was calculated using the $\mathrm{COM_{body}}$ position at three initial symmetric standing configurations: (1) legs at 90° to horizontal (2) legs at 45° to horizontal and (3) legs at 0° to horizontal, a fictitious boundary case. d was highest in the third case (0.2341m) and so the $\beta_{\max}$ was computed at d ranging from 0.23m to 0.7m, 0.7m being two times the distance of $\mathrm{COM_{body}}$ from hip joint. The response of $\beta_{\max}$ to the change in d for all three BWS strategies is shown in Figure 4. The magnitude of $\beta_{\max}$ is highest typically around the position of the $\mathrm{COM_{HAT}}$ for CF and TS strategies, while it does not show a consistent behaviour for the CW strategy. Thus, the $\mathrm{COM_{HAT}}$ was chosen as the point of application of the unloading force since it is a well-defined point and leads to high $\beta_{\max}$ values.

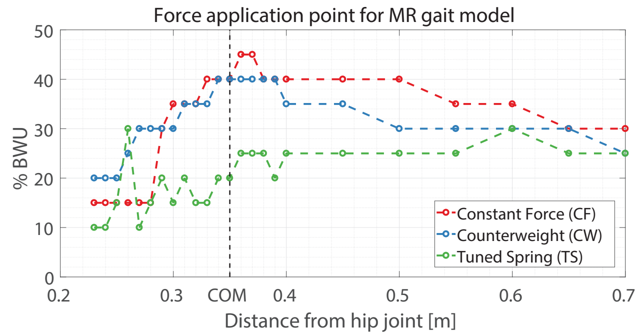

Fig. 4. Outcome of an intermediate study of the MR model. Behaviour of maximum feasible BWU ($\beta_{\max}$) with respect to the unloading force application point is shown for the three gait strategies

D. Equations of motion for the simulation

Modified equations of motion (EOM) for each model (Figures 1 and 2) are presented here. The equations which are not affected by the BWS system are not presented and can be found in the original literature.

1) Simplest walking model: Modified EOM of the Simplest walking (SW) model [11] are given below. θ represents the stance leg angle w.r.t. the vertical and ϕ is swing leg angle w.r.t. the stance leg. Following the original paper, time is scaled by $\sqrt{\frac{l}{g}}$ for the CW and TS strategies and by $\sqrt{\frac{l}{(1-u)g}}$ for the CF BWS strategy. k_{f} is the dimensionless torsional stiffness of the hip spring used for actuation [12]. The 'foot' (mass = m) is assumed to be much smaller than the 'body' (mass = M)

$$m/M \approx 0$$

.

Constant force: A term representing the constant vertical unloading force ($F_u = uMg$) can be added to the original equations [11]. Alternately, the term for gravity g can be replaced by the term $(1-u)g$ for simulated reduced gravity. The second approach was used here.

$$\ddot{\theta} = (1-u)\sin\theta \tag{4}$$
$$\ddot{\phi} = \ddot{\theta} + \dot{\theta}^2\sin\phi - (1-u)\cos\theta\sin\phi - k_{\mathrm{f}}\phi. \tag{5}$$

Counterweight system: Mass of the counterweight is uM, where M is the mass at the hip.

$$\ddot{\theta} = \frac{1-u}{1+u}\sin\theta \tag{6}$$
$$\ddot{\phi} = \ddot{\theta} + \dot{\theta}^2\sin\phi - \cos\theta\sin\phi \tag{7}$$
$$+\frac{2u}{1+u}\sin\theta\cos\phi - k_{\mathrm{f}}\phi. \tag{8}$$

Tuned spring system: Considering equations (1-3) in section II, $y_{\mathrm{c}} = l\cos\theta$ to be the vertical position of point A at time t and $y_{\mathrm{c}0} = l$ at $t = 0$,

$$\ddot{\theta} = (1-u)\sin\theta + \frac{l}{g}\omega^2 u(1-\cos\theta)\sin\theta \tag{9}$$
$$\ddot{\phi} = \ddot{\theta} + \dot{\theta}^2\sin\phi - \cos\theta\sin\phi$$
$$-u\left(1 + \frac{l}{g}\omega^2(1-\cos\theta)\right)\sin\theta\cos\phi - k_{\mathrm{f}}\phi. \tag{10}$$

2) Bipedal spring-loaded inverted pendulum model: The gait cycle in the SLIP model given in the original paper [13] is divided into three stages – initial single limb stance (SLS) of the left leg, intermittent double-limb stance (DLS) and final single limb stance (SLS) of the right cycle. The equations for horizontal acceleration do not change since BWS is assumed to influence only the vertical motion. The modified EOM for the vertical motion of the COM are presented below, where y_c represents the vertical position of the COM of the body and x is the horizontal position. The terms P and Q are the same as those defined in [13]

$$P = k(\frac{l_0}{\sqrt{x^2 + y_c^2}} - 1) \quad \& \quad Q = k(\frac{l_0}{\sqrt{(d-x)^2 + y_c^2}} - 1).$$

$d = \mathrm{FP}_{i+1,x} - \mathrm{FP}_{i,x}$ and FP is the foot point of the stance spring.

Constant force: A term representing the constant vertical unloading force ($F_u = uMg$) is added to the original equations [13].

$$\text{Initial SLS:} \quad m\ddot{y}_\mathrm{c} = Py_\mathrm{c} - m(1-u)g. \tag{11}$$
$$\text{DLS:} \quad m\ddot{y}_\mathrm{c} = Py_\mathrm{c} + Qy_\mathrm{c} - m(1-u)g. \tag{12}$$
$$\text{Final SLS:} \quad m\ddot{y}_\mathrm{c} = Qy_\mathrm{c} - m(1-u)g. \tag{13}$$

Counterweight system: Mass of the counterweight is um, where m is the mass of the body.

$$\text{Initial SLS:} \quad m\ddot{y}_\mathrm{c} = Py_\mathrm{c} - m\frac{(1-u)}{1+u}g. \tag{14}$$
$$\text{DLS:} \quad m\ddot{y}_\mathrm{c} = Py_\mathrm{c} + Qy_\mathrm{c} - m\frac{(1-u)}{1+u}g. \tag{15}$$
$$\text{Final SLS:} \quad m\ddot{y}_\mathrm{c} = Qy_\mathrm{c} - m\frac{(1-u)}{1+u}g. \tag{16}$$

Tuned spring system: Considering equations (1-3) in section II, the resulting equations for the tuned spring strategy are:

$$\text{Initial SLS:} \quad m\ddot{y}_\mathrm{c} = Py_\mathrm{c} - mg + F_\mathrm{ts}. \tag{17}$$
$$\text{DLS:} \quad m\ddot{y}_\mathrm{c} = Py_\mathrm{c} + Qy_\mathrm{c} - mg + F_\mathrm{ts}. \tag{18}$$
$$\text{Final SLS:} \quad m\ddot{y}_\mathrm{c} = Qy_\mathrm{c} - mg + F_\mathrm{ts}. \tag{19}$$

3) Muscle-reflex model: A separate Simscape sub-system was created to emulate each BWS strategy since the original Muscle-reflex gait model [14] was implemented in Simscape. The unloading force term for each BWS strategy is shown below, where m_tot is the total mass of the body and u is proportion of unloading.

Constant force:

$$F_u = um_\mathrm{tot}g. \tag{20}$$

Counterweight system: Mass of the counterweight is um_tot and $\ddot{y}_\mathrm{HAT}$ is the vertical acceleration of the upper body and the point where the counterweight BWS system is connected.

$$F_u = um_\mathrm{tot}(g - \ddot{y}_\mathrm{HAT}). \tag{21}$$

Tuned spring system: Considering $M = m_\mathrm{tot}$, y_{c0} and y_c to be the starting and current position of the $\mathrm{COM}_\mathrm{HAT}$, the unloading force term (F_u) can be obtained using equations (1-3).

$$F_u = k_\mathrm{s}(y_{\mathrm{c}0} - y_\mathrm{c} + \Delta l_0). \tag{22}$$

E. Data analysis

The four gait models were simulated for all three BWS strategies and the maximum value of percentage BWU (β_max) at which the model achieved a stable limit cycle was noted for each condition, in both optimized and non-optimized cases. Results for the BWS strategy that typically produced the highest β_max values for all gait models, were selected for the comparison with experimental data.

Relevant gait parameters data was extracted at each BWU level for all three gait models. For each BWU condition, the gait data was averaged over at least five strides in order to reduce the variability. The average step duration was considered as the inverse of cadence and the average gait cycle (stride) duration for each simulation condition was double the step duration. The proportion of each gait phase was then obtained by taking a ratio with the stride duration. The hip range of motion was calculated from the peak flexion angle following initial contact to the peak extension angle at terminal stance [42]. The knee range of motion was considered from the peak extension angle at terminal stance to the peak flexion angle at mid-swing. Peak values of the joint torques for the flexion and extension were extracted from the steady state patterns of the torques over a complete gait cycle and are indicated by negative and positive signs respectively. The two peak values for the vertical ground reaction forces (GRF) and the extrema of the anteroposterior GRF over a single gait cycle were also calculated. For muscle activity, the mean value over a complete gait cycle was considered. Since the paper [20] provides muscle activity data for individual muscles while the MR gait model typically utilizes muscle groups, the following correspondence is used for comparing the results:

TABLE I. MUSCLE GROUPS IN MR GAIT MODEL AND CORRESPONDING MUSCLES IN EXPERIMENTAL DATA

MR gait model	Experimental data
1. 'Vastus' muscle (VAS) / Quadriceps	Rectus femoris
2. Hamstring (HAM)	Biceps femoris
3. Gastrocnemius (GAS)	Lateral gastrocnemius (LG) & Medial gastrocnemius (MG)
4. Tibialis anterior (TA)	Tibialis anterior (TA)

Data for each parameter was normalized by taking a ratio with the parameter value at 0% BWU. The aim was to bring a uniformity in results and allow comparison of trends across gait models. By removing the dimensions attached to each parameter through normalizing, comparison across different gait parameters was possible. Furthermore, this normalization procedure enabled comparison with the results from the meta-analysis of experimental gait data [20]. For each gait parameter, the root mean square error (RMSE) with respect to the experimental data was calculated for overground and treadmill walking environments. The RMSE was computed and noted as a percentage of the gait parameter value at 0% BWU. The 0% BWU condition was not considered during RMSE calculation since the gait parameter data was normalized and the error at 0% BWU was always 0.

The data considered for analysis ranged from the 0% BWU condition to the highest available β_max condition. Any model

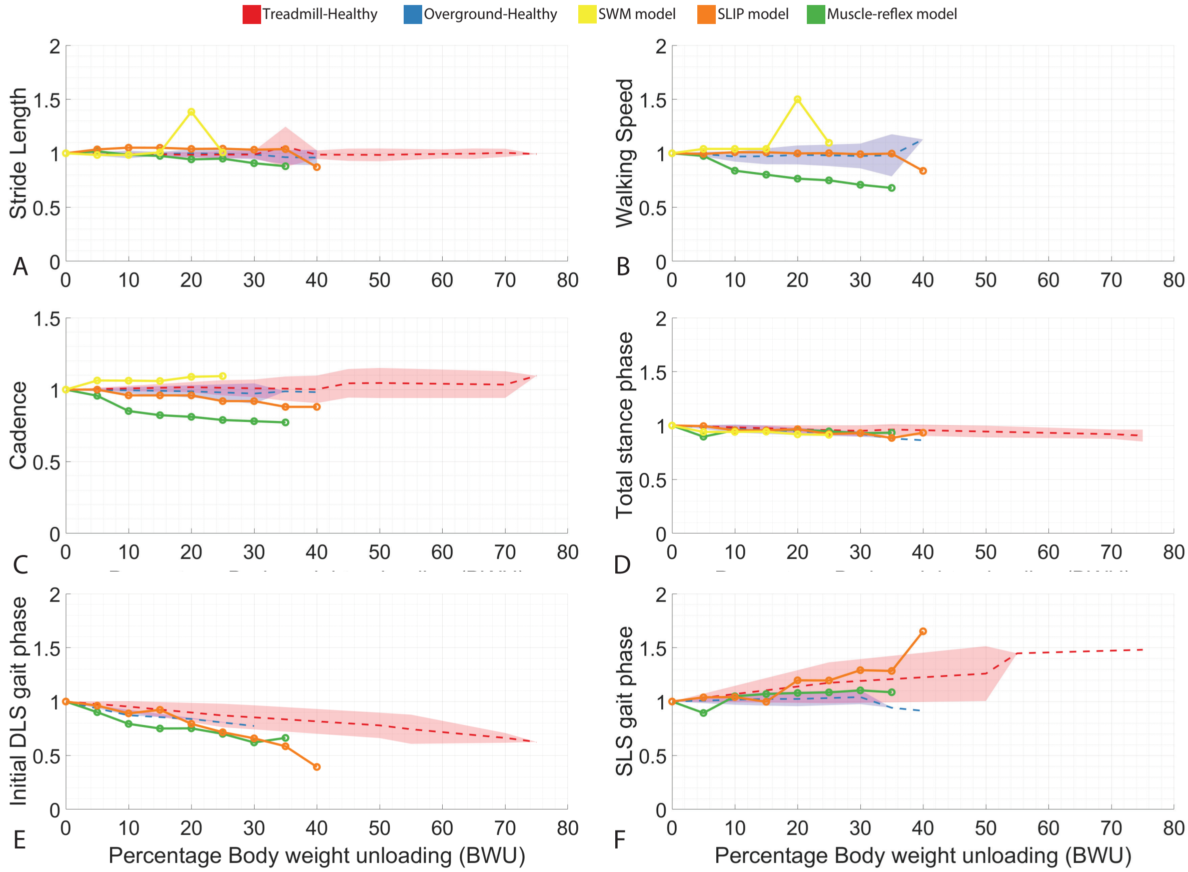

Fig. 5. Gait spatio-temporal parameters where DLS: Double limb support, SLS: Single limb support. Dashed lines represent the mean values and the shaded region represents the standard deviation for human experimental data.

having a lower $\beta_{\max}$ than the other two was penalized by assigning a value of 100% to the RMSE at the missing BWU data conditions. However, if the one of the models had a higher $\beta_{\max}$ for a BWS strategy other than the selected one, this model would have been penalized solely due to the choice of BWS strategy. To avoid influencing the analysis purely on the basis of the choice of BWS strategy, the data analysis was restricted to the highest available data point for such a model and not the overall highest $\beta_{\max}$ condition.

III. Results

The $\beta_{\max}$ values for the three models and the BWS strategies are presented in table II. The gait parameter data was available for up to 25%, 40% and 35% for the SW, SLIP and MR models respectively (table II). A RMSE value of 100% was assigned for all relevant gait parameters at the 30% and 35% BWU conditions for the SW model since this data was not available. The MR model, however, was not penalized for the lack of data at 40% since it has the highest $\beta_{\max}$ for the CF BWS strategy and the lack of data at 40% BWU was the result of choosing the TS strategy for analysis. To avoid influencing the analysis this way, the data analysis was restricted to 35% BWU instead of 40% BWU, which is the overall highest available data point.

TABLE II. Maximum value of BWU ($\beta_{\max}$) at which the model still achieved a stable gait cycle. Table shows $\beta_{\max}$ values for both optimized (Opt) and non-optimized (Non-opt) cases.

BWS strategy	**CF**		**CW**		**TS**	
Gait model	Non-opt	Opt	Non-opt	Opt	Non-opt	Opt
SW	5	15	5	5	10	25
SLIP	15	15	0	0	35	40
MR	40 (CF)	—	30	—	35	—

The gait parameter values at different levels of BWU for each gait model are plotted in Figures 5-8, along with the experimental data [20] for healthy individuals walking in overground and treadmill environments. The RMSE for each model and the relevant gait parameters are presented in table III. A lower value of the RMSE indicates a better fit with the experimental data and consequently, the better suited a gait model is for the investigation of that specific gait parameter. The comparison of gait models is based only on the RMSE values for overground condition. Values for the treadmill condition are presented only for comparison with the overground condition for the same model and not between two models. Qualitative descriptions for results of each gait model

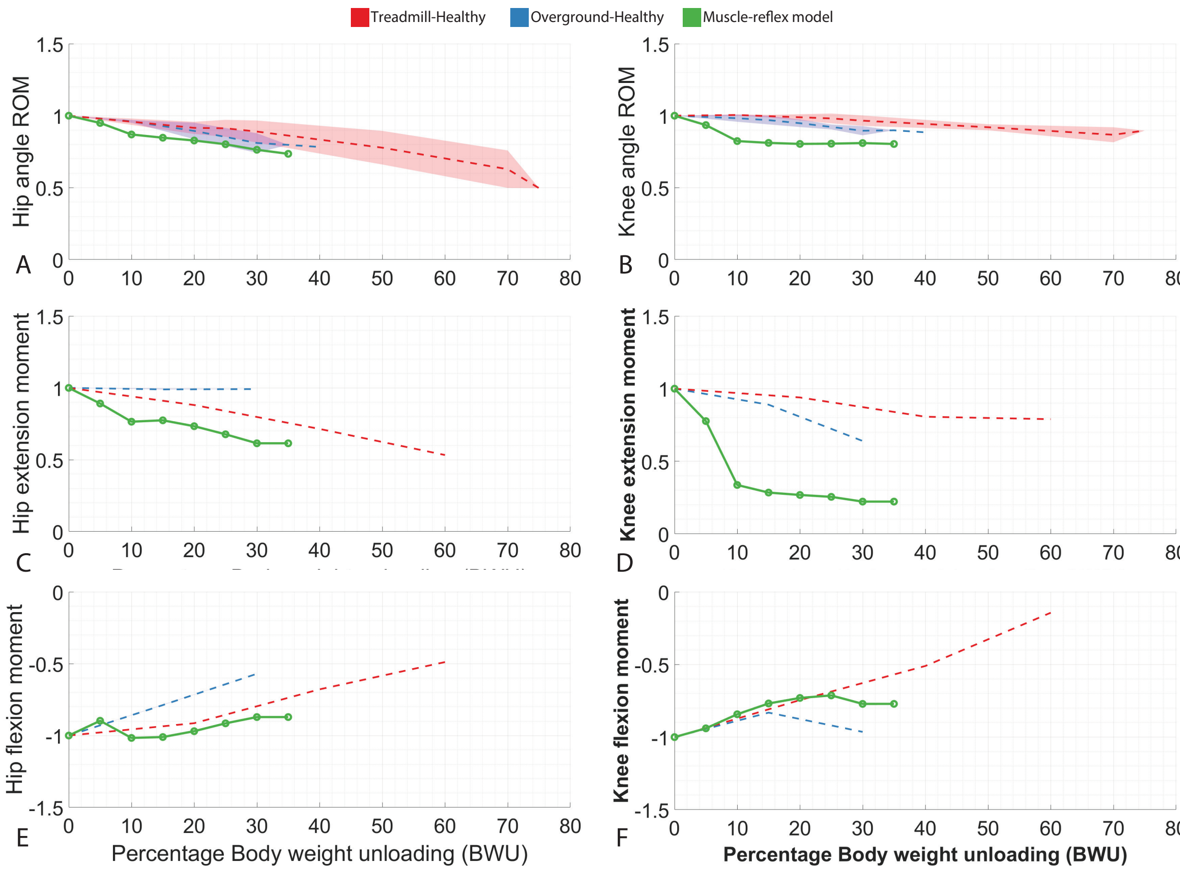

Fig. 6. Hip and knee joint dynamics where ROM: Range of motion. Dashed lines represent the mean values and the shaded region represents the standard deviation for human experimental data.

are presented below. Descriptions like ‘high’ and ‘low’ RMSE values are relative to the RMSE values for the same parameter across different gait models.

A. *Simplest walking model*

The Simplest walking (SW) model was analyzed only for the gait spatio-temporal parameters like stride length, cadence, walking speed and the total stance phase. Since the model has an instantaneous double support phase, the double support and single support phases were not considered separately. The ground reaction forces (GRF) were also not considered since they did not follow the characteristic pattern of anthropomorphic bipedal gait [43]. The SW model has the highest RMSE values for all the relevant gait parameters. For the majority of gait parameters, the SW model has lower RMSE values for overground walking as compared to treadmill walking.

B. *Spring-loaded inverted pendulum gait model*

The Spring-loaded inverted pendulum (SLIP) gait model was investigated for all gait spatio-temporal parameters and the vertical GRF. The anteroposterior GRF were not analyzed since they failed to represent the patterns from anthropomorphic data [44]. The SLIP model has the lowest RMSE values for all relevant gait characteristics (Figures 5 & 7) except for single limb stance phase and stride length. For single limb stance phase (SLS), it shows a high value. This can be observed in Figure 5.F, where the SLS value increases drastically as compared to the increase in the experimental SLS data. For stride length (SL), it shows a low RMSE value (5.39 %), comparable to the MR gait model (5.29 %). Apart from the SLS and SL gait parameters, all other gait parameters have a higher RMSE value in case of the treadmill walking condition.

C. *Muscle-reflex gait model*

The Muscle-reflex (MR) gait model was the only model which could be tested for almost all gait parameters mentioned in [20]. Except for stride length and single limb stance phase, the MR model has higher RMSE values for other gait parameters than the SLIP model but lower than the SW model (table III). Of the 23 gait parameters analyzed, this model has a RMSE of less than 10% for seven characteristics - stride length, total stance phase, single limb stance, ankle moment, the deceleration peak of anteroposterior GRF and the second peak of vertical GRF. However, it has a high RMSE for knee extension moment (Figure 6.E). Furthermore, for 7 gait parameters, the MR model shows lesser RMSE values for the

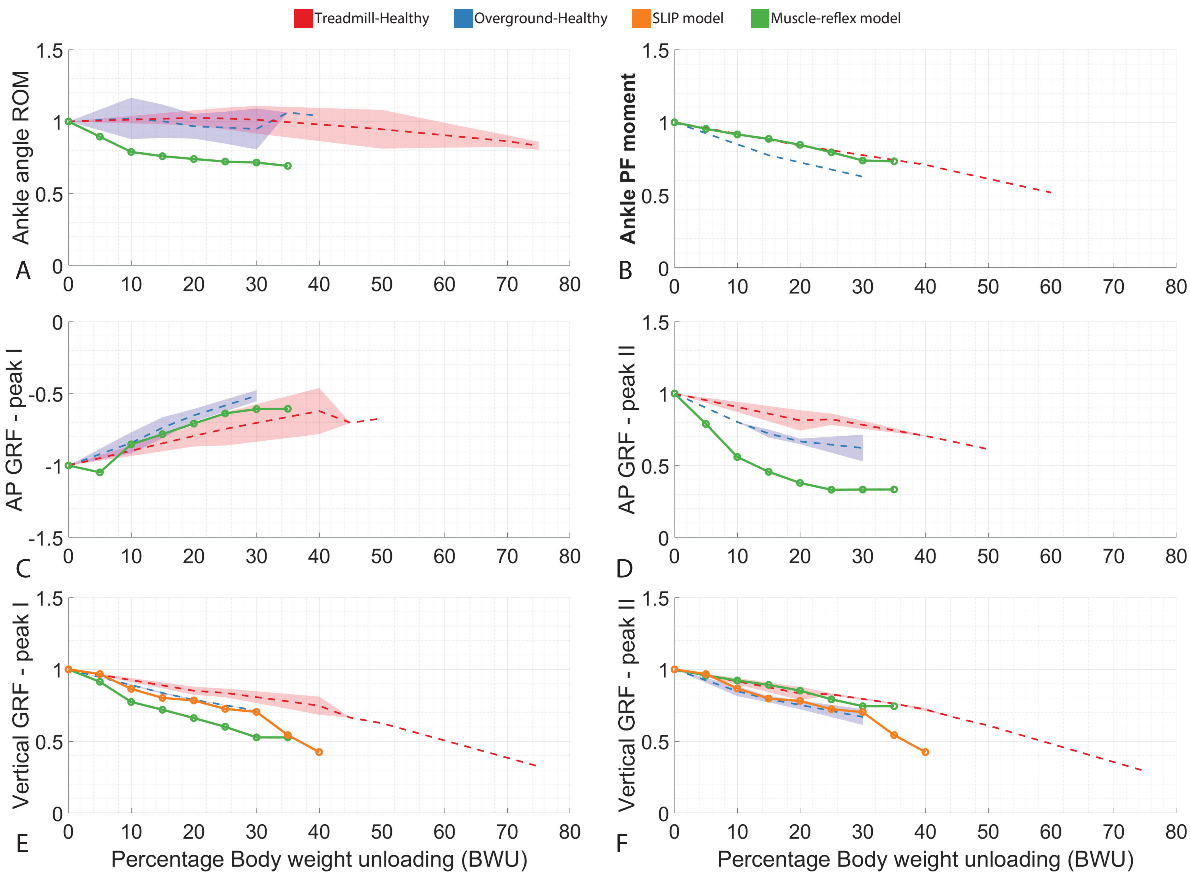

Fig. 7. Ankle joint dynamics and ground reaction forces where PF: plantarflexion, GRF: ground reaction forces and AP: anteroposterior. Dashed lines represent the mean values and the shaded region represents the standard deviation for human experimental data.

treadmill walking condition than the overground one. These parameters include all joint moments except knee extension moment (Figures 6 & 7), the second peak of vertical GRF and the muscle activity for lateral gastrocnemius and tibialis anterior muscles (Figure 8).

IV. Discussion

This section details the implications of our results for the gait models, followed by a brief discussion on the different BWS strategies and concludes with a description of the limitations of our research.

A. *Influence of BWU on simulated gait*

The three gait models show a stronger influence of BWU on most gait parameters than the experimental human data for both treadmill and overground walking conditions (Figures 5-8). While the human data presents a higher influence of BWU on the dynamic gait characteristics than the gait spatio-temporal parameters and joint angles, the gait models present a larger effect for joint angles, the single limb support (SLS) and double limb support (DLS) phases. This is reflected in the higher RMSE for the SLS and DLS phases and the joint angle ROM, as compared to the RMSE values for other spatio-temporal parameters. The stride length and walking speed plots (Figure 5.A-B) for the SW model show a distinct peak at 20% BWU. Since the initial conditions were optimized for each BWU level, this step length value is one of the multiple solutions (set of initial conditions) for which the SW model achieved a stable gait cycle at 20% BWU. However, since the repetitions of the optimization process yielded a similarly high step length value, this result could merit further investigation into the SW model. The fast walking speed at 20% BWU is the consequence of the longer step length at the same condition. The SLIP model presents a larger increase in the proportion of SLS phase relative to human data and the MR gait model, which leads to a high RMSE. This phenomenon can be attributed to the stabilization effect of the unloading force during SLS. This effect is more pronounced in the SLIP model than MR model, which is comparatively more robust to disturbances [14].

In case of the MR model, the unloading force produces an additional torque about the hip joint which needs to be counter-balanced by the hip joint. This leads to an increase in the hip flexion moment and a decrease in the hip extension moment and subsequently affects the knee joint moments as well (Figure 6.C-F). While analyzing the data, it was noted that the peak knee extension torque shifted temporally from

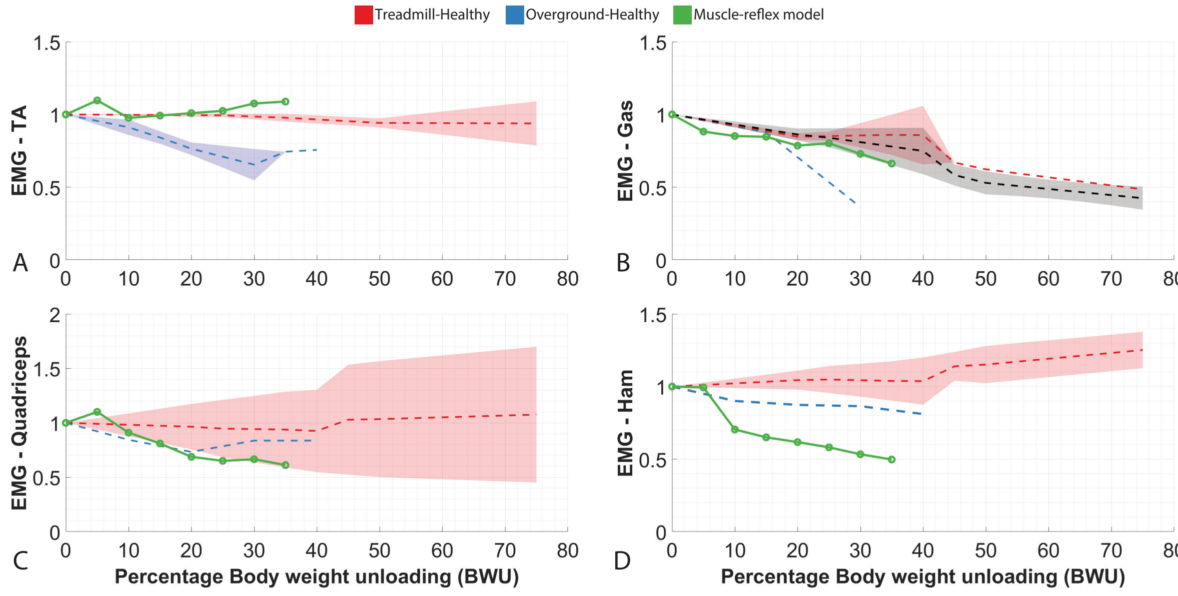

Fig. 8. Mean muscle activity over gait cycle where TA: tibialis anterior, Gas: Gastrocnemius and Ham: Hamstring. In subfigure B, black and red lines indicate the muscle activity in lateral and medial gastrocnemius respectively (only treadmill walking data). Dashed lines represent the mean values and the shaded region represents the standard deviation for human experimental data.

just after initial contact to just before toe-off at 10% BWU. This temporal change in torque peak led to a sharp drop in KME magnitude, as seen in Figure 6.E. This could explain the sizable deviations from the human data for the hip and knee joint moments and thus the high RMSE. While the ankle plantarflexion moment in the MR model is less affected by BWU than in humans, ankle angle ROM drops almost 20% lower than the human data. This reduction in ankle ROM should lead to a higher reduction in the forward push-off force (anteroposterior GRF peak II) and a lower reduction in the vertical push-off force (vertical GRF peak II) as compared to human data. As predicted, this effect can indeed be seen in the GRF (Figure 7). In case of the muscle activities, muscle groups in the MR gait model were compared to individual muscles in the experimental data (table I). While the muscle activities of the individual muscles are correlated to the muscle groups [45], the MR model shows high RMSE values ($> 10\%$) for all muscles except the Lateral Gastrocnemius.

In case of the TS BWS strategy, the range of $\beta_{\max}$ values in case of all three gait models lies between 25% to 40% (table II. Above this range, the unloading force leads to such a strong influence on the gait parameters that the gait models are unable to attain a stable gait. This range of $\beta_{\max}$ values aligns closely to the 30% BWU level, up to which the influence of BWU on gait spatio-temporal and kinematic parameters has been shown to be limited [10, 20, 46–50]. The gait models follow similar qualitative trends (increasing/decreasing) as the human data, despite high RMSE values for some parameters like the joint moments and muscle activities. This indicates that the response of gait models to BWU is akin to that of humans, albeit slightly exaggerated. These results thus strengthen the idea that reasonably simple gait models can be effectively used to simulate the effects of BWU on human locomotion.

B. Comparison of gait models

The SW model shows the highest RMSE values for all four gait parameters (table III). While Figure 5 does not reflect such high RMSE values ($\approx 50\%$), these values are expected due to the penalization process explained in the earlier section (II.D). The MR Model, despite being the most complex model in this selection, performs worse than the SLIP model. Typically, the aim of modulated BWU is to enable the magnitude of spatio-temporal parameters to be similar to the values during unsupported walking or to retain the m-shape of the vertical GRF [21, 23]. These gait parameters are present in the SLIP model, which enables their use in both, controlling the unloading force and investigating their response to the BWU. Thus, the SLIP model can be used effectively to simulate the effects of modulated BWU on human gait.

The muscle reflexes and initial conditions for the MR model were not optimized for BWU, which might partially explain its lower performance. While an optimization would have led to a higher value of $\beta_{\max}$, the non-optimized model still yields comparatively high $\beta_{\max}$ values (table II). However, hand-tuning the model to suit every modulated BWU level would require extensive human data from experiments with modulated BWU and obtaining this data is difficult unless more modulated BWS systems are designed. Since the main idea behind simulating gait models is to use them for the design of modulated BWU strategies, hand-tuning of the MR model using experimental data available a priori is not feasible. Even if such data is available, the large number of tunable parameters makes it difficult to optimize the MR gait model. The MR model could still be useful in certain scenarios, wherein

TABLE III. SUMMARY OF THE RESULTS OF GAIT PARAMETERS FOR THE THREE GAIT MODELS. ROOT MEAN SQUARE ERROR VALUES WITH RESPECT TO THE EXPERIMENTAL DATA FOR OVERGROUND (OG) AND TREADMILL (TM) ENVIRONMENTS ARE PRESENTED HERE.

Root mean squared error values						
Experimental data	**Overground (OG)**			**Treadmill (TM)**		
Gait model	**SW**	**SLIP**	**MR**	**SW**	**SLIP**	**MR**
Gait parameter	%	%	%	%	%	%
01. Stride length	55.40	5.39	5.29	55.54	4.94	7.84
02. Cadence	53.94	5.48	17.03	53.73	7.65	19.41
03. Walking speed	57.21	2.41	21.07		—	
04. Gait phases - Stance	53.51	1.24	4.06	53.58	3.59	4.09
05. Gait phases - Double limb stance	–	6.95	10.05	–	14.13	16.77
06. Gait phases - Single limb stance	–	18.43	8.18	–	6.73	8.89
07. Hip joint - ROM	–	–	6.42	–	–	9.93
08. Knee joint - ROM	–	–	12.19	–	–	16.21
09. Ankle joint - ROM	–	–	24.67	–	–	26.25
10. Hip extension moment	–	–	26.35	–	–	14.99
11. Hip flexion moment	–	–	22.56	–	–	8.12
12. Knee extension moment	–	–	48.87	–	–	60.63
13. Knee flexion moment	–	–	13.36	–	–	9.64
14. Ankle plantarflexion moment	–	–	9.93	–	–	1.57
15. Anteroposterior GRF peak - I	–	–	7.40	–	–	8.23
16. Anteroposterior GRF peak - II	–	–	26.05	–	–	39.82
17. Vertical GRF peak - I	–	2.28	13.03	–	11.58	20.26
18. Vertical GRF peak - II	–	2.69	8.02	–	10.58	2.61
19. Muscle activity - Quadriceps	–	–	13.97	–	–	23.63
20. Muscle activity - Hamstrings	–	–	25.94	–	–	41.44
21. Muscle activity - Medial Gastrocnemius	–	–	–	–	–	7.85
22. Muscle activity - Lateral Gastrocnemius	–	–	19.30	–	–	10.30
23. Muscle activity - Tibialis anterior	–	–	26.78	–	–	6.71

the muscle-reflexes could be tuned to emulate the pathological muscle function in individuals with neuromuscular disorders.

C. Nature of the unloading force

The SW and SLIP models responded in a considerably different manner to the three BWS strategies and to a lesser extent, the MR model did as well. Table II shows that the $\beta_{\max}$ values were typically highest for the TS BWS strategy and lowest for the CW BWS strategy. The TS BWS strategy is based on the hypothesis that tuning the stiffness of a spring-based BWS according to the cadence of the user leads to a more transparent BWS system and a gait which is more similar to unsupported walking [29]. The $\beta_{\max}$ values for the TS BWS strategy support the above hypothesis and make a case for a detailed comparison of actual gait parameter trends across the three BWS strategies for a specific gait model. While the major BWS systems currently used are based on a constant unloading force strategy, these results indicate a shift towards TS BWS strategy might be beneficial.

The $\beta_{\max}$ values for the TS and CF BWS strategies were highest if the unloading force was applied close to the COM of the upper body (Figure 4). This suggests that the unloading force, even at a small distance from upper body COM, leads to a destabilizing moment about the upper body. An in-depth investigation of the behavior of gait characteristics at different locations of the BWU application point could be useful for the design of harness systems and for making a choice between pelvic or body harness-based attachment systems. In case of the CW BWS strategy, the COM location changes due to the counterweight, thus making it difficult to predict the $\beta_{\max}$ behaviour.

D. Limitations of this study

Cost of transport (COT) or the energy cost for walking was not evaluated while comparing the gait models. Literature states that COT decreases with the increase in BWU and that COT is an important governing factor for gait transitions [20]. Mechanical work can be calculated from the joint power consumption and while it is correlated to the COT, it cannot be used to accurately determine the COT [51].

Gait models based on an optimization approach [52, 53], gait models governed by neural control and central pattern generators [54–56] and OpenSim-based gait models [57] were not considered in this study. However, the initial selection of five gait models covers most of the main features of human gait like mechanical stability, compliant nature of legs, segmented legs, muscle-reflex architecture, m-shape of vertical GRF and upper body balance control [44, 58].

V. CONCLUSION

The goal of this research was to benchmark gait models based on their suitability to the simulation of human walking with body weight support. Gait models were simulated under the influence of Constant force, Counterweight and Tuned spring BWS strategies. The outcome for the Tuned spring BWS strategy raises doubts over the dominant idea of aiming towards constant unloading force and merits further investigation into the use of a Tuned spring approach. The results of this work demonstrate the usefulness of gait models for BWS simulation and suggest the SLIP model to be more suitable for BWS simulations than the Simplest Walker and the Muscle-reflex models.

LIST OF ABBREVIATIONS

BWS: Body weight support
BWU: Body weight unloading
BWST: Body weight supported training
SW: Simplest walking
SLIP: Spring-loaded inverted pendulum
MR: Muscle-reflex
CF: Constant force
CW: Counterweight
TS: Tuned spring
TM: Treadmill
OG: Overground
ROM: Range of motion
PF: Plantarflexion GRF: Ground reaction forces
AP: Anteroposterior
EMG: Electromyography
ST: Stance phase
DLS: Double limb support
SLS: Single limb support
MG: Medial gastrocnemius
LG: Lateral gastrocnemius
TA: Tibialis anterior
Gas: Gastrocnemius
Ham: Hamstring

ACKNOWLEDGMENT

We would like to thank the authors of the original gait models and the authors of the Matlab implementations of these models. This work is (partially) funded by Eurostars Project 10 152 â LEAP and Marie Curie career integration grant.

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Appendix

The Appendix provides supplementary information and additional results with the aim of improving the understanding of main paper. Results for the capture-point control-based Virtual pivot point gait model are provided in chapter 1. Only the results for relevant gait parameters are presented and in the same format as the main paper. Chapter 2 introduces the gait parameters used for analysis in the paper and explains their physiological origin and relevance. In addition to this, it presents a brief description of a typical experiment designed to investigate the influence of body weight support on human walking. Furthermore, this chapter also motivates the classification of body weight support systems based on the nature of the unloading force. Chapter 3 contains the research paper from which the human experimental data was extracted. Finally, chapter 4 presents some background information for the three gait models and their simulation process.

CONTENTS

1

ADDITIONAL RESULTS - THE VPP MODEL

The Virtual pivot point (VPP) gait model was not included in the main paper since the available implementation of the VPP model was based on Capture-point control instead of the Constant angle-of-attack controller used in the original model. These results are presented in the current chapter.

1.1. RESPONSE TO BODY WEIGHT UNLOADING

The simulation procedure is described in section II of the main paper. All four gait models were simulated with three body weight support (BWS) strategies, Simulated gravity (SG), Counterweight (CW) and Tuned spring (TS). The body weight unloading (BWU) level was increased from 0% to 100% in 5% steps. For the VPP model, the unloading force was applied at the center of mass of the upper body (COM_{body}). At every BWU level, the model parameters and initial conditions were optimized to achieve a stable limit cycle and consequently the longest walking distance. Results for the maximum feasible BWU value (β_{max}) i.e the BWU value for which the model could achieve a stable gait cycle, are shown in Table 1.1.

Table 1.1: Maximum value of BWU (β_{max}) at which the model still achieved a stable gait cycle. Table shows β_{max} values for both optimized (opt) and non-optimized (non-opt) cases.

BWS strategy	**SG**		**CW**		**TS**	
Gait model	Non-opt	Opt	Non-opt	Opt	Non-opt	Opt
SW	5	15	5	5	10	25
SLIP	15	15	0	0	35	40
VPP	25	25	0	0	0	0
MR	40 (CF)	—	30	—	35	—

Since the TS BWS strategy typically resulted in the highest β_{max} values for the Simplest Walking (SW), Spring-loaded inverted pendulum (SLIP) and Muscle-reflex (MR) models, results for the TS BWS strategy were used for the analysis of gait characteristics. However, results of the SG BWS strategy were selected for the VPP model as it could not achieve a stable gait for other two strategies (Table 1.1). This is another reason why the model was not considered in the benchmarking study.

1.2. ANALYSIS OF GAIT PARAMETERS

The same process, described in section II of the paper, was used for analyzing the gait data generated from the VPP model. The VPP model was investigated for all gait spatio-temporal parameters, hip joint torques and the anteroposterior and vertical GRF (Figure 1.1). The response for all gait parameters, except the gait phases, is qualitatively (increasing/decreasing) similar for the VPP, other gait models and human experimental data. However, the gait phases show opposite trends in case of the VPP model. The single limb support (SLS) phase decreases while the initial double limb support (DLS) phase and the total stance phase increase in the VPP data.

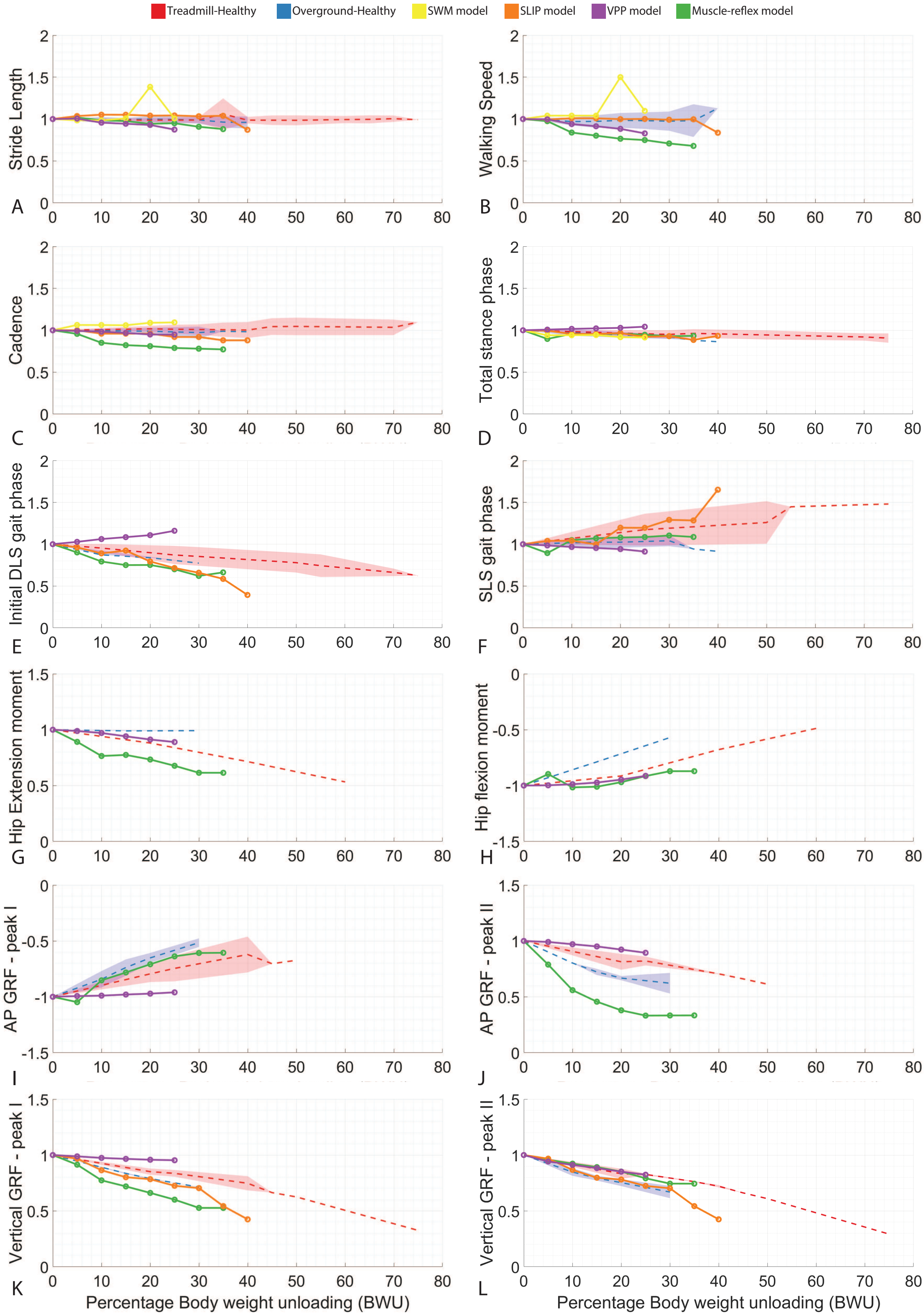


Figure 1.1: Gait spatio-temporal parameters where DLS: Double limb support, SLS: Single limb support, hip joint moments and ground reaction forces where GRF: ground reaction forces and AP: anteroposterior

For each gait parameter, the root mean square error (RMSE) with respect to the experimental data was calculated for overground and treadmill walking environments. The RMSE, as a percentage of the gait parameter value at 0% BWU, is reported in Table 1.2. The gait parameter data was available for up to 25% for the SW and VPP models, and 40% and 35% for the SLIP and MR models respectively (Table 1.1). The SW model and the VPP models were penalized for having a lower β_{max} as compared to the other two models. A RMSE value of 100% was assigned for all relevant gait parameters at the 30% and 35% BWU conditions since this data was not available from the simulation. These RMSE values were used as a basis for comparison of the VPP model to other models. Along with the SW model, the VPP model has the highest RMSE values for relevant gait parameters. While Figure 1.1 does not reflect such high RMSE numbers ($\approx$ 50%), these values are expected due to the penalization process. The unexpected behaviour of the gait phases, coupled with the high RMSE values, make the Capture-point control based VPP model unsuitable for simulating the influence of body weight support on human gait.

Table 1.2: Summary of the results of gait parameters for the four gait models. Root mean square error values with respect to the experimental data for overground (OG) and treadmill (TM) environments are presented here.

Root mean squared error values								
Experimental data	**Overground (OG)**				**Treadmill (TM)**			
Gait model	**SW**	**SLIP**	**VPP**	**MR**	**SW**	**SLIP**	**VPP**	**MR**
Gait parameter	%	%	%	%	%	%	%	%
1. Stride length	55.40	5.39	53.76	5.29	55.54	4.94	53.71	7.84
2. Cadence	53.94	5.48	53.50	17.03	53.73	7.65	53.61	19.41
3. Walking speed	57.21	2.41	53.95	21.07		—		
4. Gait phases - Stance	53.51	1.24	53.84	4.06	53.58	3.59	55.34	4.09
5. Gait phases - Double limb stance	–	6.95	57.26	10.05	–	14.13	55.64	16.77
6. Gait phases - Single limb stance	–	18.43	53.79	8.18	–	6.73	53.65	8.89
7. Hip extension moment	–	–	53.71	26.35	–	–	53.53	14.99
8. Hip flexion moment	–	–	55.84	22.56	–	–	53.55	8.12
9. Anteroposterior GRF peak - I	–	–	57.69	7.40	–	–	54.85	8.23
10. Anteroposterior GRF peak - II	–	–	56.27	26.05	–	–	53.87	39.82
11. Vertical GRF peak - I	–	2.28	54.71	13.03	–	11.58	53.91	20.26
12. Vertical GRF peak - II	–	2.69	53.92	8.02	–	10.58	53.46	2.61

2

Background for gait analysis

This chapter details the parameters used to study and characterize human gait in experiments, especially for the experiments concerning BWS systems. These parameters are identified by surveying literature and then grouped together based on the nature of these parameters (Table 2.1). A typical BWS-based experimental setup used to study these gait characteristics is also described in this chapter, followed by a motivation for the classification of BWS systems.

Group	Parameters
Gait spatio-temporal parameters	Stride length, walking speed, cadence, gait phases
Joint kinematics	Hip, knee and ankle joint range of motion (ROM)
Gait dynamics	Joint moments
Ground reaction forces	Anteroposterior and vertical GRF
Muscle activity	EMG signals

Table 2.1: Categorization of gait characteristics

2.1. Gait spatio-temporal parameters

Stride length

The book 'Biomechanics and Motor Control of Human Movement' [1] has the following definition for stride length, "*It is the horizontal distance covered along the plane of progression during one stride; it is the distance covered from IC (instantaneous centre) to IC of the same foot, equal to the sum of the two step lengths and will be equal for left and right limbs if the person is walking in a straight line, even in the presence of marked asymmetry. Specific step lengths for right and left side must be measured within the same stride.*" While step lengths can be different for the two sides (left and right), especially if one of them is paretic, the stride length for them would be similar if the person is walking in a straight line. Analysis of step width was deemed beyond the scope of this work.

Walking speed

Mean walking speed can be defined as the ratio of the distance traveled by the participant and the total time taken. This distance can be determined by measuring the position of a marker close to the center of mass of the body like the greater trochanter or the sacral marker. Apart from walking speed, another measure is the stride speed, which is the ratio of stride length and stride duration. However, walking speed is a more widely used parameter, which makes it possible to compare the results across multiple studies. While the data from the overground experiments shows that the walking speed changes with an increase in the amount of unloading [2–6], treadmills usually force the participants to walk at a constant speed, thus imposing unnatural gait dynamics.

Cadence

Cadence or gait frequency is defined as the number of steps taken per unit time (generally 1 second). It can be calculated by taking the ratio of the number of steps taken while walking a fixed distance and the time needed to cross that distance [3]. Usually, there is negligible difference in cadence for overground and treadmill walking [7].

Gait phases

The gait cycle (GC) can be defined as the interval of time between any of the periodic events of walking, like initial contact of foot with ground till the same foot contacts the ground again. The gait cycle is comprised of two phases, stance (when the concerned leg provides support) and swing (Figure 2.1). The stance phase can be subdivided into 3 parts, including initial double limb support DLS, single limb support SLS, and terminal double limb support DLS. The initial and terminal DLS phases typically account for 10% of the gait cycle, while SLS typically accounts for 40% (total stance phase is approximately 60% total) [8]. The swing phase for this same limb makes for the remaining 40% of the cycle. Ipsilateral (concerned leg) swing temporally corresponds to single stance by the contralateral (other leg) limb. During double stance, the two legs do not generally share equal load and slight variations occur in the percentage of stance and swing phases [8].

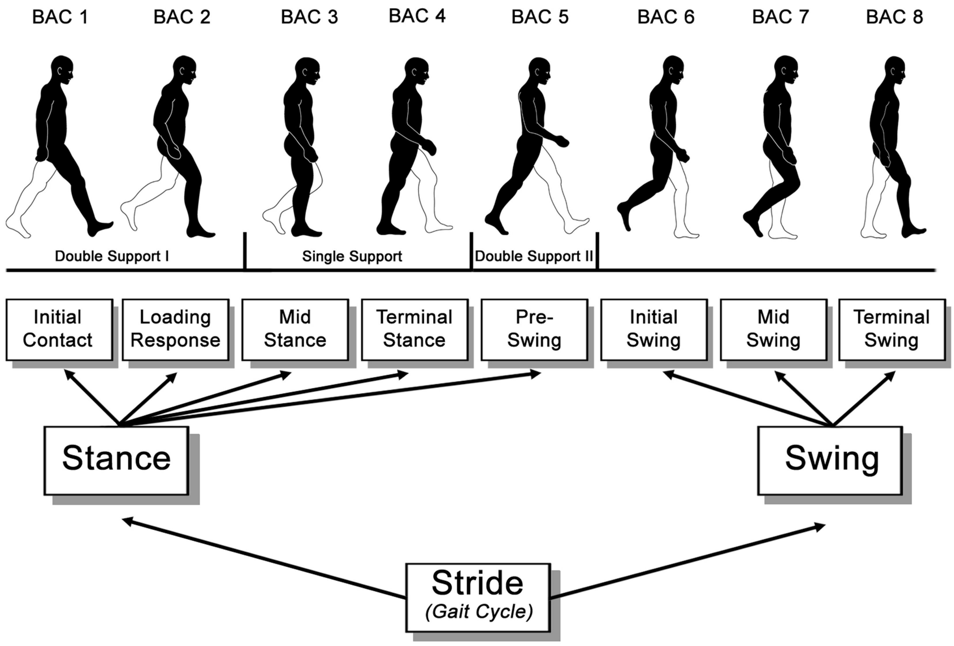


Figure 2.1: Phases of gait cycle. Figure reprinted with permission from "The mental representation of the human gait in young and older adults", T. Stöckel et al., Frontiers in psychology. Copyright 2015 Frontiers Media S.A. [9]

Stance phase is comprised of five phases — initial contact, loading response, mid stance, terminal stance and pre-swing (Figure 2.1). The first two phases, initial contact and loading response (0-10% GC) exist during initial double support. Initial contact phase happens with the transfer of weight onto the new stance phase leg while preserving gait speed, maintaining stability and attenuation of impact force on foot. SLS phase involves the movement of the CoM of the body over the supporting limb. This starts with mid stance (10-30% GC) and continues through terminal stance (30-50% GC). This phase includes heel rise of the support foot and terminates with contralateral foot contact. [8]. The final stance element, preswing, begins with terminal double support and ends with toe-off of the ipsilateral leg. The swing phase includes foot clearance and advancing of the trailing leg and consists of three phases, initial swing (60-73% GC), mid swing (73-87% GC), and terminal swing (87-100% GC) [8].

A gait cycle (GC) is equivalent to a stride since the duration of a stride is the interval between sequential initial floor contacts by the same limb. Following this, the gait cycle time and the gait cycle parameters (for eg. SLS) are calculated as:

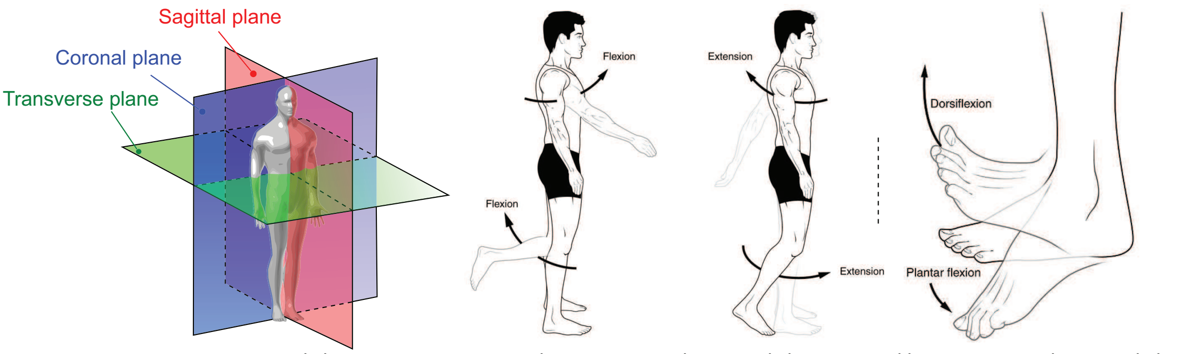
$$GC(s) = \frac{2}{cadence(HZ)}, \quad SLS(\%) = \frac{SLS(s)}{GC(s)}$$

2.2. JOINT KINEMATICS AND KINETICS

This section introduces the anatomical terms describing the joint kinematics of the legs followed by a description of the motion of hip, knee and ankle joints respectively.

ANATOMICAL DEFINITIONS

The motion of lower limb joints in 3D space is generally decomposed into 2D trajectories in three different planes for ease of understanding and analysis. These three planes are known as sagittal, coronal and transverse planes (Figure 2.2) [1]. The sagittal plane is a vertical plane running from front to back and it divides the body or any of its parts into right and left sides. The coronal plane is a vertical plane running from side to side and it separates the body or any of its parts into anterior (front) and posterior (back) portions. The transverse plane is typically a horizontal plane, parallel to the ground and running through the center of the body. It divides the body into upper and lower parts.



a. Anatomical Planes b. Movement in the Sagittal Plane c. Ankle movement in the Sagittal plane

Figure 2.2: Anatomical planes and movements. Figure 'a' adapted from [10], 'b' and 'c' are reprinted from OpenStax CNX 2017 [11].

Gait analysis experiments based on joint kinematics usually consider only the movements in sagittal plane since the extent of motion of the legs in this plane is significantly higher than that in other two planes [1]. The movements that take place in sagittal plane are flexion and extension and they involve anterior or posterior movements of legs(Figure 2.2b). Extension straightens the joint by increasing the angle between two bones at a joint while flexion decreases it (bending). For the hip joint, flexion is bringing the thigh forward and upward while moving the thigh posterior is extension. At the knee joint, extension involves straightening the knee while flexion is the bending of the knee in posterior direction. For the ankle joint, lifting the front of the foot towards the anterior leg is dorsiflexion, while pointing the toes downward is plantar flexion.

HIP JOINT KINEMATICS

Hip joint kinematics has been measured and analyzed in literature using multiple measures like hip angle at initial contact (IC), maximum hip extension angle during stance, hip angle at preswing, hip angle at toe off, maximum hip flexion angle during swing and the range of motion (ROM) for hip joint [12], usually specified in degrees. ROM of the hip joint is usually specified as the sum of the maximum flexion and extension angles. Range of Motion (ROM) is used as a metric for comparison in this report since it is the most recurrently used measure in literature.

KNEE JOINT KINEMATICS

Knee joint kinematics is generally specified in gait analysis literature using parameters like knee extension angle at initial contact (IC), peak knee flexion angle during early stance, knee angle at preswing, knee angle at toe-off, peak knee flexion angle during swing and knee angle range of motion (ROM) [12], generally measured in degrees. In spite of so many parameters to choose from, due to its widespread use, range of motion (ROM) is used as a parameter for comparison across literature. However, unlike hip joint, ROM for knee joint is given by the difference between peak flexion angle (during swing) and peak extension angle (at initial contact) [4].

ANKLE JOINT KINEMATICS

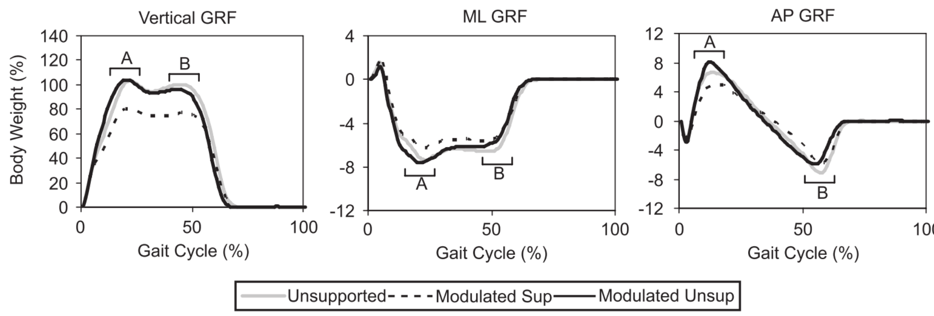
Ankle joint kinematics has been examined in concerned literature using parameters like ankle angle at initial contact, maximum value of ankle angle during early stance, ankle angle at preswing (dorsiflexion), ankle angle at toe-off (TO), maximum plantarflexion during swing, ankle angular velocity and ankle range of motion (ROM) [12, 13]. Ankle range of motion (ROM) is the yardstick used for comparison of results across literature in this work. This ankle ROM is calculated as the sum of maximum dorsiflexion and plantarflexion angles.

Joint moments

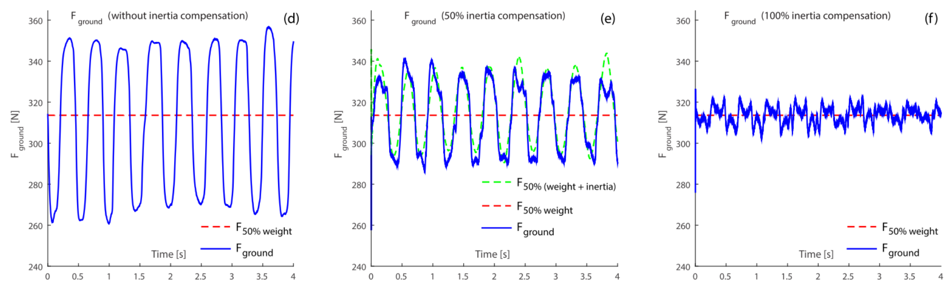
The concerned research considers joint moments in the sagittal plane. These joint moments are generally calculated using inverse dynamics method and then normalized according to body mass and height for each participant [13]. The peak extension and flexion moments for hip are generally measured during loading response and toe-off respectively while the peak moments for knee joint are typically observed during the loading response and terminal stance [4, 14]. In case of the ankle joint, only the plantarflexion moment is considered since it plays a critical role in the push-off phase.

Ground reaction forces

Ground reaction forces (GRF) can be divided into three main components, the vertical, mediolateral (ML) and the anteroposterior (AP) ground reaction forces. Typical GRF curves along a gait cycle for all three components can be seen in Figure 2.3c. Generally, the vertical peaks of GRF plot occur during the loading response and the terminal stance phases and are caused due to the vertical motion of the body center of mass. The AP GRF indicate the acceleration and deceleration for during push-off and heel-strike. Only the magnitude of the two vertical GRF peaks and both positive and negative peaks of the AP GRF is analyzed in this work.



(a) Variation with modulated BWU - vertical, lateral & anteroposterior GRF



(b) GRF for a modulated BWS system which compensates for inertial forces

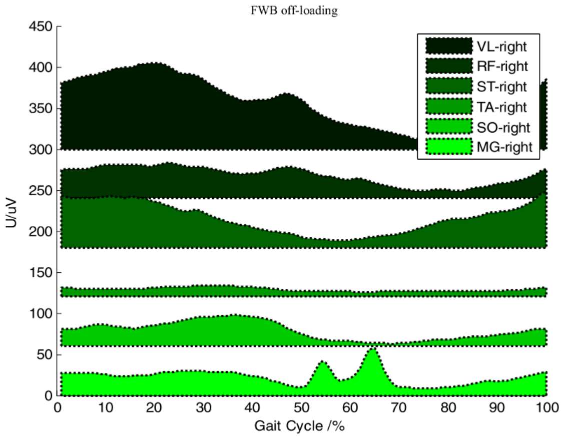
Figure 2.3: Variation of ground reaction forces (GRF) with BWU level for constant and modulated BWS system. Magnitude of GRF (V, AP & ML) decreases with the increase in BWU level. Figure 'a' reprinted with permission from "Gait synchronized force modulation during the stance period of one limb achieved by an active partial body weight support system", J. R. Franz et al., Journal of biomechanics. Copyright 2008 Elsevier. Figure 'b' reprinted with permission from "Gravity-assist: A series elastic body weight support system with inertia compensation", H.Munawar and V. Patoglu, Intelligent Robots and Systems (IROS). Copyright 2016 IEEE. [15, 16].

2.3. Muscle activity

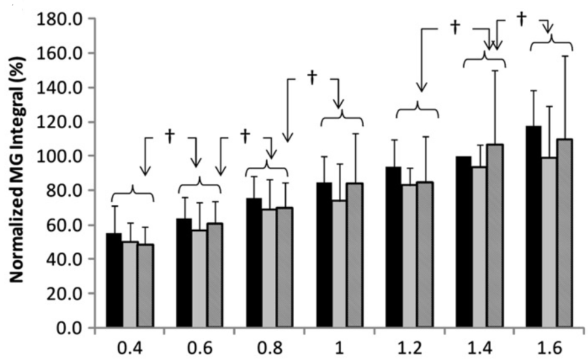
Muscle activity is quantified and recorded through the technique of electromyography (EMG). EMG is performed using an instrument called an electromyograph, which records the electric potential produced by muscle cells when they are neurally or electrically activated [17]. Typical muscle activation patterns during a gait cycle can be seen in Figure 2.4 and a typical setup for measuring EMG signals can be seen in Figure 2.5. EMG signals are recorded using electrodes attached to the skin surface and the muscles generally analyzed are the tibialis anterior (TA), soleus (SO), vastus lateralis (VL), vastus medialis (VM), lateral gastrocnemius

(LG), gastrocnemius medialis (MG), gastrocnemius(GA), bicep femoris (BF), rectus femoris (RF), semitendinosus/semimembranous (ST), gluteus maximus (GMax), gluteus medius (GMed), and adductor longus (AL) and gluteus maximus (GM) [6, 6, 18, 19]. The activity patterns of antagonistic muscle pairs (RF, VL - BF and LG - TA) are organized approximately in a reciprocal manner.

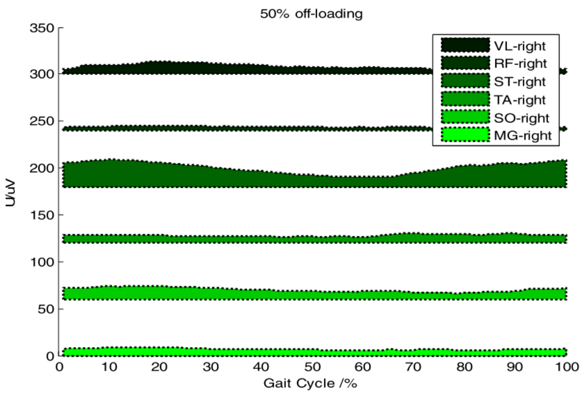
Apart from using EMG to investigate effect of BWU on muscle activation patternss, EMG measurement can also be used to investigate the influence of BWU on the nocioceptive flexion reflex (NFR) [20]. NFR is a muscle withdrawal reflex initiated as a response to the activation of pain signaling nerve fibers. EMG is used to monitor the activity in the BF muscle while applying electrical stimulation to the lower leg (on the same side) and the individual reports onset of pain when the NFR is aroused. EMG signals are also used to study the modulation of H-reflex in the solues muscle [21]. Soleus H-reflex examination is a standard method for understanding the alterations produced in the excitability patterns due to neurological impairments.



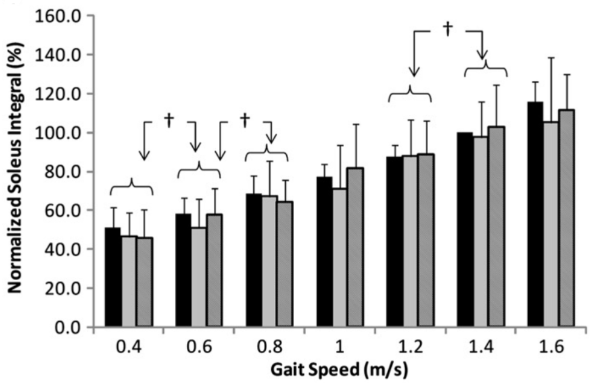
(a) Muscle activation pattern at 0%BWU



(b) Medial gastrocnemius (MG) muscle activity



(c) 50% BWU



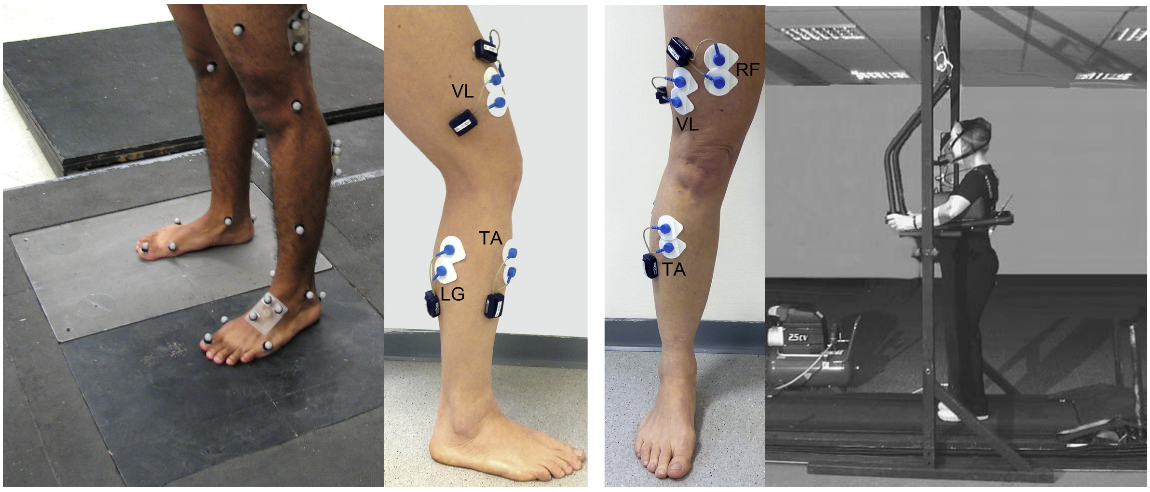
(d) Soleus (So) muscle activity

Figure 2.4: The muscle activation patterns generally become flatter with the increase in BWU level. Figures 'a', 'c' and 'e' adapted with permission from "Modulation of weight off-loading level over body-weight supported locomotion training", P. Wang et al., IEEE International Conference Rehabilitation Robotics(ICORR). Copyright 2011 IEEE. Figures 'b','d' and 'f' adapted with permission from "The influence of body weight support on ankle mechanics during treadmill walking", M. D. Lewek, Journal of biomechanics. Copyright Elsevier 2011. [13, 19].

2.4. Typical experimental setup based on BWS systems

During a typical BWS system-based experiment, participants are mechanically supported in an adjustable full body harness, attached to the BWS system which can be active or passive in nature. The connection between the harness and the BWS system usually comprises of a load cell to measure the amount of weight unloaded by the BWS system. After wearing the harness and being connected to the BWS system, the participants are allowed to familiarize with the conditions and walking environment (overground/treadmill) through practice trials. The walking environment is often embedded with force plates for measuring the ground reaction forces

(GRF) produced by the participants.



(a) Passive reflective markers for motion capture & force plate embedded in walking environment

(b) EMG electrodes for recording muscle activity

(c) VO2 measurement setup

Figure 2.5: Preparation of participants for experiment — markers and EMG. Figure 'a' is adapted from Wikimedia Commons library [22]. Figure 'b' reprinted with permission from "Effects of body weight unloading on electromyographic activity during overground walking" by A. G. Fischer et al., Journal of Electromyography and Kinesiology. Copyright 2015 Elsevier. Figure 'c' reprinted with permission from "Physiological costs and temporo-spatial parameters of walking on a treadmill vary with body weight unloading and speed in both healthy young and older women", E. E. Thomas et al., European journal of applied physiology. Copyright 2007 Springer. [18, 23]

For an experiment based on motion capture, reflective markers (usually passive) are placed at key anatomical locations or landmarks (Figure 2.5a), examples of those being the lateral malleolus, greater trochanter, head of the fifth metatarsal, lateral epicondyle of the femur, acromion etc., which are necessary to define the foot, shank, thigh, and trunk segments, respectively. For a full body motion capture, additional markers are placed on the pelvis and the trunk. The locations of these markers and subsequently their trajectory is recorded through the use of infrared cameras. These trajectories are later subjected to kinematic and dynamic analyses to determine the behaviour of key gait parameters. Software packages such as OpenSim can be used for these analyses [24]. EMG activity in the leg muscles is recorded by placing two surface EMG electrodes on the belly of the leg muscles (Figure 2.5b) and the commonly measured muscles include, Rectus Femoris (RF), Lateral Gastrocnemius (LG), Tibialis Anterior (TA), Vastus Lateralis (VL) etc. The force plates, motion capture system and the EMG recording are sufficient to measure most of the outcome measures discussed in this section. Apart from these three setups, VO2 (volume of oxygen intake) can also be measured in order to estimate the metabolic energy consumption of the participants.

2.5. Classification of BWS systems

Kang et al. [25] undertook a survey of the preferences of the medical community towards robotic therapy. According to their results, the two most preferred designs are the treadmill type (47.5%) and the over-ground walking type (40%), followed by "foot plate-based gait trainer" (11.5%), and "fixed-gait trainer" (1%) [25]. BWS systems form a key component of both, the treadmill and the overground walking type of robotic treatment designs. Neurological and/or physical impairments often prevent patients from supporting their own weight during walking. Not only do the BWS systems enable walking movement by reducing the load that directly supported by the patient, they also ensure safety and stability of the patient. BWS systems typically consist of a cable which supports a harness worn by the patient. This cable applies a uni-directional vertical unloading force to the patient through a system of pulleys and a counterpoise. This counterpoise can be actuated or passive, based on springs or a counterweight (see Figure 2.6). The amount of unloading provided by the BWS system is generally high at the beginning of the treatment and is gradually decreased as the patient improves his/her ability to support his/her weight. Afferent signals to the brain increase by gradually decreasing unloading provided by the BWS which supports gait 'retraining' through appropriate activation of sensory receptors at the appropriate times during the gait cycle. For this report, the main factors considered are the

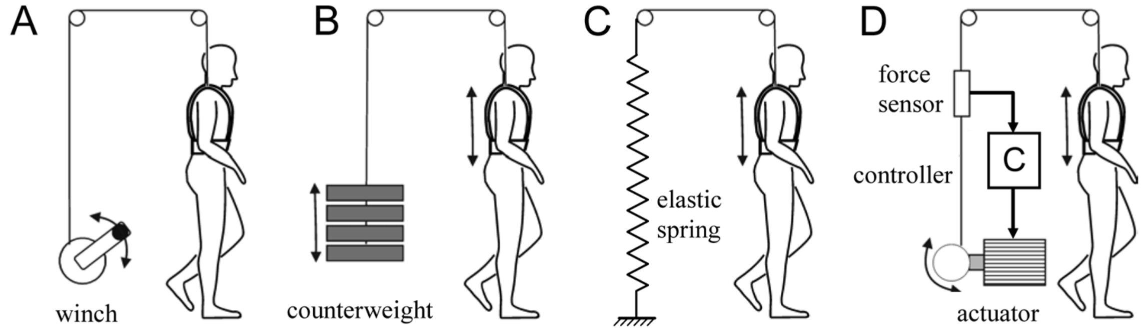


Figure 2.6: Body weight support systems concepts: (A) Static BWS (B) Passive dynamic BWS with adjustable counterweights (C) Elastic element-based passive dynamic BWS (D) Active dynamic BWS. Reprinted with permission from "A novel mechatronic body weight support system," by M. Frey et al., 2006, IEEE Transactions on Neural Systems and Rehabilitation Engineering. Copyright 2006 IEEE [27]

manner in which the BWS systems provide unloading force (unmodulated or modulated force) and the walking environment for the patient. Other factors that can be important are the workspace of the BWS sytem (2D/3D), attachment to the patient (pelvic/body harness), etc. Thus, in this report, BWS systems are broadly classified into two main types - unmodulated body weight unloading and modulated BWU systems, the details of which are mentioned in the section below. Regarding the walking environment used during treatment, the two most commonly used ones are a treadmill and plain ground or an overground (fixed ground) rehabilitation system as it is known in literature.

Classification based on unloading force

The BWS systems are classified into two types, unmodulated and modulated BWU systems, based on whether they are designed to modulate the unloading force to compensate for the inertial forces produced due to the unloaded mass of the patient while walking under body weight unloading.

Unmodulated BWU

The main design goal for a constant BWU system is to produce a desired amount of constant unloading force on the user. These systems can be active or passive, depending on the presence or absence of actuators (motors usually). Frey et al. classify these systems into three main types, the static BWS, passive dynamic BWS and elastic-element based passive dynamic BWS. The static BWS system (Figure 2.6A) is comprised of a mechanism, usually a motor-actuated winch, that is calibrated to unload a predefined amount of weight, when the patient is connected to the BWS system [26]. However, a static BWS system cannot ensure consistent percentage of body weight unloading due to the vertical movement of the body center of mass (CoM) while walking. As the body moves downwards the harness becomes tighter, thus restricting pelvis movement. Thus, static BWS systems might be less effective for gait rehabilitation due to their unsuitability to natural gait.

Passive dynamic BWS system based on counterweight (Figure 2.6B) utilizes an adjustable counterweight to dynamically provide a specific percentage of body weight unloading [27]. Since the counterweight moves vertically in response to the vertical movement of the CoM of the patients, it can maintain a constant static unloading force. However, the vertical movement of the counterweight generates additional inertial forces, which cause large fluctuations in the support forces. To overcome the problem of inertial forces due to the counterweight, elastic elements like springs can be used to provide body weight unloading through the tension in the spring [27]. Passive elastic BWS systems are designed using this principle (Figure 2.6C). However, since the amount of unloading provided by a passive elastic element varies with its length, the percentage of unloading changes with the vertical movement of the CoM of the patient as he or she walks. Percentage of body weight unloading used during nascent stages of therapy can be very large (up to 60%) to compensate for the patient's weakness [27]. However, with such a high percentage of unloading, the passive BWS devices can cause large dynamic loads on patients, especially in case of counterweight based devices.

Modulated BWU

Human body behaves as an inertial load whilst being supported by the BWS system. The patient's weight results in a continuous static load on the BWS system while the vertical motion of the center of mass (CoM) of the body results in inertial forces, proportional to the vertical acceleration of the CoM. High magnitude of such undesired inertial forces can lead to deviations from the desired level of loading, thus hampering the intended operation of the BWS system. In addition to this, the inertia of the counterweight used in the BWS or the weight of the exoskeleton supported by the BWS can produce a decrease in the natural gait frequency, thus hindering the rehabilitation process [28].

An effective body weight unloading technique should compensate for the inertial forces resulting from the unloaded mass and the BWS system, so that the patient perceives a constant weight unloading, even under dynamic movements like walking[27, 29]. This strategy avoids an improper ratio between the weight and inertial forces rendered to the patient and has the potential to promote natural gait. A modulated BWU system can monitor the interaction forces between the patient and the BWS system and control the actuator responsible for weight unloading so as to achieve a constant vertical force on the patient [16, 29–31]. Unloading force can also be modulated based on the phases of the gait cycle or trajectory of the centre of pressure [32]. Such a actively controlled BWS system can provide the patient with a feeling of dynamically reduced mass, as if part of his/her body mass is removed.

Classification based on walking environment

The nature of the walking environment is important, since it is a crucial factor for facilitating the skill transfer to everyday movements [33]. Not only do the the requirements for walking on treadmill from walking overground vary in terms of balance control and propulsion, the walking speed chosen on a treadmill is not typically self-selected unlike overground gait [2].

Treadmill gait training

Treadmill-based BWST systems generally comprise a body weight support system which is attached to the patients while walking/training on a treadmill. During treadmill-based gait training, the patient's legs are sometimes attached to the frame of the exoskeleton using straps and the exoskeleton provides powered assistance during movement. A typical example of a treadmill based exoskeleton rehabilitation device is the Lokomat [34]. Lokomat works on a position control based paradigm where the patient (wearer) follows the movement (gait) produced by the device and cannot make adjustments to the movement. A study found that use of Lokomat produced slightly better results than conventional physiotherapy [35]. However, newer exoskeleton based gait training robots like LOPES [36] and ALEX [37] provide force field based assistance to the patient and the provided assistance can be varied according to the phases of the treatment. Patients who are able to move their legs without assistance, are trained directly using a BWS system on a treadmill.

Overground gait training

Overground gait rehabilitation systems comprise of the robots that servo-follow the patient's walking motions over ground or overhead of the patient. Patients can move and walk under their own control using overground gait trainers rather than being forced through walk through a predefined motion trajectory. One example of an overground gait training rehabilitation robot is the KineAssist device, developed by Kinea Design, LLC, for gait and balance training [38]. KineAssist consists of a torso and pelvis harness supported by a mobile robotic base which is controlled according to the contact forces detected in the pelvic support. Other examples are the Multidirectional transparent support for overground gait training or FLOAT device [30], and the ZeroG system [31] which consist of an overhead cable-based weight supporting system. These devices are aimed at the rehabilitation of patients who are able to move their legs without assistance.

3

Meta-analysis of BWS experiments

The main paper is based on the results from a meta-analysis of experimental data for humans walking with body weight support. Since the research paper presenting this meta-analysis is currently under review in the Journal of Neuroengineering and Rehabilitation (JNER), a copy of this paper is included here. The meta-analysis was conducted as a part of the literature study and is based on the literature study report submitted earlier [39].

RESEARCH

Influence of body weight unloading on human gait characteristics: a systematic review

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Abstract

Background: Body weight support (BWS) systems have shown promise as rehabilitation tools for neurologically impaired individuals. This paper reviews the experiment-based research on BWS systems with the aim: (1) To investigate the influence of body weight unloading (BWU) on gait characteristics; (2) To study whether the effects of BWS differ between treadmill and overground walking and (3) To investigate if modulated BWU influences gait characteristics less than unmodulated BWU

Method: A systematic literature search was conducted in the following search engines: Pubmed, Scopus, Web of Science and Google Scholar. Statistical analysis was used to quantify the effects of BWU on gait parameters.

Results: 55 studies of experiments with healthy and neurologically impaired individuals walking in a BWS system were included and 38 of these were used for the statistical analysis. Literature was classified using three distinctions: (1) treadmill or overground walking; (2) the type of subjects and (3) the nature of unloading force. Only 27% studies were based on neurologically impaired subjects; a low number considering that they are the primary user group for BWS systems. The studies included BWU from 5% to 100 % and the 30% and 50% BWU conditions were the most widely studied. The number of participants varied from 1 to 28, with an average of 12. It was seen that due to the increase in BWU level, joint moments, muscle activity, energy cost of walking and ground reaction forces (GRF) showed higher reduction compared to gait spatio-temporal and joint kinematic parameters. The influence of BWU on kinematic and spatio-temporal gait parameters appeared to be limited up to 30% unloading. 5 gait characteristics presented different behavior in response to BWU for overground and treadmill walking. Remaining 21 gait characteristics showed similar behavior but different magnitude of change for overground and treadmill walking. Modulated unloading force generally led to less difference from the 0% condition than unmodulated unloading.

Conclusion: This review has shown that BWU influences all gait characteristics, albeit with important differences between the kinematic, spatio-temporal and kinetic characteristics. BWU showed stronger influence on the kinetic characteristics of gait than on the spatio-temporal parameters and the kinematic characteristics. It was ascertained that treadmill and overground walking can alter the effects of BWU in a different manner. Our results indicate that task-specific gait training is likely to be achievable at a BWU level of 30% and below.

Keywords: Body weight support; Rehabilitation; Gait characteristics

Background

Body weight supported training (BWST) has shown promise in providing improvements in motor function, locomotion ability and balance in patients suffering from damage to the nervous system [1–6]. Example patient groups are spinal cord injury (SCI) patients, stroke patients and Parkinson's disease patients. During BWST, a certain percentage of the patient's body weight is supported by an overhead suspension system through a harness worn by the patient [7]. BWS systems enable physiotherapists to assess and correct gait patterns during interventions, without the obligation of providing full physical assistance [8]. In one of the earliest studies on this subject, Wernig et al. discovered that, with body weight supported treadmill training (BWSTT) for around 7 months, SCI patients having complete or partial paralysis could learn to perform voluntary bipedal stepping with joint stabilization and body weight bearing [9]. Patients with a paralyzed limb were able to walk short distances while bearing their own weight and in absence of joint stabilizers like knee braces. Recently, with the advent of robotic rehabilitation devices, the total duration of training and its precision can be increased even more without increasing the burden on the physiotherapists, thus enabling wider application of BWST [10, 11].

A BWS system is typically composed of an apparatus in which the patient is mechanically supported by a harness while walking on a treadmill or overground [12]. The constraints and support provided by the BWS system helps the subjects' vertical alignment and stability of the trunk throughout ambulation [12, 13]. This, in turn, can provide neurologically impaired users the confidence to start rehabilitation early after surgery or trauma to regain balance and locomotion without the fear of a fall [8]. Furthermore, BWS systems also allow perturbation of patients in a safe environment in order for the patients to improve their balance. BWS decreases lower-extremity load, thus facilitating step initiation [14]. When a treadmill is used, the treadmill can aid hip extension in the stance leg, critical to the initiation of swing phase, and supply temporal cues associated with stepping [15]. Although it is still debated [16], several studies indicate that task specificity in rehabilitation training is crucial [17]. BWST makes use of such task-oriented outlook with the aim of improving the performance of that task. Further benefits of BWST seem to be improved cardiovascular health, increased glucose tolerance and insulin sensitivity, growth in muscle mass, reduction in visceral fat and enhanced psychological well-being [18].

The theoretical underpinning of BWST as a clinical intervention is the concept of neuroplasticity [19]. The purpose of BWST is to supply the injured nervous system with necessary and appropriate sensory input signals for stimulating the intact spinal cord networks in order to facilitate their continued involvement even when supraspinal input is undermined [20]. Barbeau et al. first suggested the use of a treadmill and BWS for the gait rehabilitation of patients with SCI [21]. Since the study by Barbeau et al., other studies have reinforced the idea that BWST of persons with clinically complete or incomplete SCI induces functional re-organization of spinal neuronal networks, which leads to improvements in motor function and decreased muscle co-contractions [10, 19, 22–26].

It is still an open question how to choose a suitable level of body weight unloading (BWU). BWU is defined here as the percentage of a patient's body weight that is unloaded by the BWS system. Oftentimes, the selection of a particular BWU level has been based largely on a subjective judgment, such as what percentage of unloading leads to seemingly normal gait. For example, in one study, the maximum BWU level was chosen based on whether the participant can still maintain heel contact and achieve toe push-off [5]. It is known from research on motor control that a particular task determines what movements are needed, and even small adaptations of tasks may affect movement strategy [27]. Thus, while planning therapeutic interventions, it is necessary to consider the adaptation of walking patterns that result from BWU [28]. Task-specific training which leads to the preservation of natural gait characteristics under BWU, may improve the outcome of treatment [29, 30]. Therefore, a vast amount of research is focused on the effect of BWU on gait characteristics. Some research is also dedicated to non-constant, modulated BWU, an example being modulation of unloading force based on the gait phase [31]. However, the researchers study different gait parameters and the results are not always consistent. Therefore, the aim of this study is to combine all studies about the influence of BWU on the gait in order to answer the central question investigated in this paper:

`'How does body weight unloading affect gait characteristics?'`

This question is divided into three sub-goals: (1) To quantify and analyze the influence of body weight unloading (BWU) on gait characteristics; (2) To study whether the effects of BWS differ between treadmill and overground walking and (3) To investigate if modulated BWU influences gait characteristics less than unmodulated BWU. The scope of the literature covered in this paper is limited to the research published from 1991 to 2016 and includes studies about walking in both neurologically impaired adults and adults with no known motor disorders. Though BWST is also utilized for pediatric rehabilitation [32–35], a combined meta-analysis of studies with adults and children as subjects would make it difficult to interpret the results. Consequently, the scope of this review is confined to experiments with adult participants. The pathologies included in this review are spinal cord injury, cerebrovascular accident (stroke) and Parkinson's disease. While literature about the clinical outcomes of BWST in adults with other neuromuscular disorders is available [36, 37], studies about the influence of body weight unloading on gait biomechanics during training are missing. Besides rehabilitation, BWS systems have been used to study the effects of reduced gravity on gait for the purpose of space exploration [38–45]. The study by Richter et al. reviewed this literature and hence it was not analyzed in this paper [46]. However, a comparison of results with the paper by Richter et al. is presented in the discussion section. Comparison is also provided with the results by Harvill et al. for locomotion at lunar gravity [47], since these were not covered in the review by Richter et al.

This paper is divided into three sections. The first section explains the methodology pursued while conducting the literature review. This is followed by a detailed description of the parameters used to study effects of BWU on gait and the results and trends for each of these parameters reported in existing experimental research. The paper concludes with a discussion on the important gait outcome measures studied in literature, the distinction between results for treadmill-based and overground studies and a overview of the experiments aimed at investigating effects of body weight unloading (BWU) on gait.

Method

Search strategy

Identification of potentially relevant literature was conducted through electronic search of four databases: Pubmed, Scopus, Web of Science and Google Scholar. The following search terms were utilized using the Boolean mode - (weight support OR weight unloading OR simulated gravity OR reduced gravity) AND (body OR gait OR locomotion OR characteristics OR rehabilitation OR overground OR treadmill OR spinal cord injury OR stroke OR parkinson's OR walking). Searches were limited to studies based on adult human subjects performing a walking task, published in English language and up to the year 2016. These search results were extended by examining the references lists of returned articles. Apart from these searches, citations of the papers presenting the design of electromechanical body weight support systems were explored [31, 48–52]. Literature about the effects of water immersion on human gait is not considered relevant due to the drag and damping produced by the viscosity of water [42, 53].

Literature identification

The population of interest were both healthy individuals and individuals suffering from neurological disorders like SCI, stroke and Parkinson's disease. Though the symptoms and effects of these disorders are different, all the concerned patients can use BWST for rehabilitation. The relevant outcome measures were all gait characteristics including kinetic, kinematic and spatiotemporal parameters along with energy consumption, ground reaction forces and muscle activity.

For an article to be included in this review, the source article had to describe: (1) whether an electromechanical BWS system was used; (2) nature of weight unloading; (3) treadmill/overground walking; (4) gait characteristics used and (5) a gait analysis experiment with at least one participant. The last criterion excludes any simulation-based studies. Despite inclusion of any particular study in this review, the data from that study was excluded from the meta-analysis if: (1) the experiment involved less than five participants (2) data of the clinical outcome of BWS training was presented instead of the data showing the influence of BWU on gait during body weight supported walking.

Gait data is excluded from the studies where effects of each BWU level are tested at different speeds and the studies in which only the change in gait parameters is mentioned [54–56]. Results of the experiments where assistive devices were used in combination with a BWS systems are not incorporated in the analysis [57–61]. Since provision of guidance through assistive devices can lead to a lower muscle activity and these effects can dominate over the influence of BWU [62], exclusion of the data from these studies improves the reliability of the statistical analysis. One paper presented data in the form of a linear regression instead of providing raw data [63]. As this might lead to misleading values of the coefficient of determination (R^2), this data is also ruled out from the statistical analysis.

Results are also not included from the experiments featuring a BWS system with a tilted walkway [40, 41, 64], nearly-parabolic flight [38, 65], partial immersion in water [66], horizontal suspension systems [45], saddle-based body attachment [67–69] and air-pressure unloading force around lower body [70–72]. These different types of BWS systems might influence the gait differently than the more widely used harness-based vertical BWS systems [46, 73] and thus their exclusion from the meta-analysis.

Data extraction

The following data was extracted from the selected literature: (1) BWS type; (2) treadmill or overground walking; (3) participants' physiological condition (neurologically impaired or otherwise); (4) number of participants; (5) unloading conditions tested for and (6) gait characteristics investigated and their units of measurement (see Additional file 1). Mean values for each independent gait parameter were obtained from the studies.

Study classification

The literature was classified based on three distinctions. First, treadmill and overground studies were distinguished. This nature of the walking environment is important, since it has been claimed to be a critical factor for facilitating the skill transfer to everyday movements [74]. For example, the walking speed chosen on a treadmill is typically not self-selected unlike overground gait [12]. In addition, a body-weight support system above a treadmill also provides relative assistance for propulsion, while the same does not necessarily hold for overground gait [75]. The training outcomes for treadmill-based and overground training might also be different. Field-fote et al. discovered that walking distance improved to a larger degree with overground training as compared to treadmill-based training for individuals with chronic motor incomplete SCI [76]. Second, healthy subject studies and studies with neurologically impaired subjects were differentiated. One could also distinguish between different patient groups, but due to the small amount of studies per patient group, it was decided to categorize all neurologically impaired subjects together. Finally, there is a distinction between constant and modulated BWU systems based on whether or not they are designed to modulate the unloading force.

The subject results were classified into six categories, based on the first two distinctions: (1) treadmill-healthy (TH); (2) overground-healthy (OH); (3) treadmill-patients (TP) with results for both legs considered together and (4) overground-patients with results for both legs considered together (OP). The last two categories were further divided into results for (5) non-paretic leg (TPN and OPN) and (6) the paretic leg (TPP and OPP). The outcomes for these groups will be shown throughout the results section.

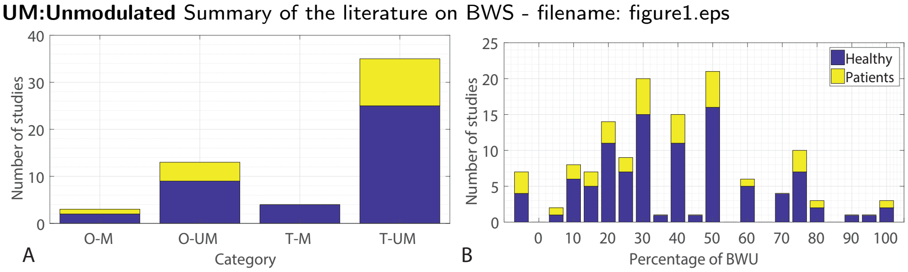
The types of BWS systems were grouped into four different groups, based on the first and third distinction: (1) treadmill modulated (T-M); (2) overground modulated (O-M); (3) treadmill unmodulated (T-UM) and (4) overground unmodulated (O-UM). The difference in the outcomes between these groups is examined in the discussion section.

Figure 1 shows the number of studies found per category and the amount of BWU studied by them. Table 1 presents the studies as classified per type of BWS that they use.

Table 1 Classification of BWST literature based on nature of unloading force and walking environment. Research based on patients is printed in bold. Research carried out on neurologically impaired participants is usually a comparison between patients and healthy subjects. * - two different publications by the same author/s in the same year. Studies by David et al. and Delussu et al. were conducted on GaitTrainer, a stepping plate based device [57, 59].

	Treadmill-based	Overground
Modulated BWU	Franz 2007 [31], Franz 2008 [77], Van Thuc 2015 [78], Munawar 2016 [79] – *4 studies*	Morbi 2012 [80], Fenuta 2014 [60], **Fenuta 2014*** [81] - 3 studies
Constant BWU	Finch 1991 [54], Farley 1992 [68], Donelan 1997 [67], Kram 1997 [69], **Dietz 1997** [106], **Harkema 1997** [23], **Dietz 1998** [107], Colby 1999 [101], **Hesse 1999** [94], Stephens 1999 [87], Griffin 1999 [100], **Danielsson 2000** [102], Ferris 2001 [85], Ivanenko 2002 [88], Threlkeld 2003 [93], Grabowski 2005 [103], **Ferris 2004** [104], Van Hedel 2006 [89], **David 2006** [57], **Phadke 2007** [25], Thomas 2007 [90], McGowan 2008 [99], Aaslund 2008 [27], McGowan 2009 [86], Lewek 2010 [83], Klarner 2010 [58], Kuno 2012 [63], Goldberg 2013 [97], **Delussu 2014** [59], **Meyns 2014** [122], Van Kammen 2014 [92], Worthen-Chaudhari 2015 [56], Swinnen 2015 [61], Dragunas 2016 [13], Van Kammen 2016 [62] – *35 studies*	Patino 2007 [28], **Sousa 2009** [12], **Burgess 2010** [95], Wang 2011 [105], Serrao 2012 [84], Barela 2014 [98], Fischer 2015 [8], Fischer 2015* [108], **Hurt 2015** [55], **Jung 2016** [82], Fischer 2016 [109], Mun 2016 [96], Ye 2016 [124] – *13 studies*

Figure 1 Summary of BWS studies where O:Overground, M:Modulated, T:Treadmill, UM:Unmodulated Summary of the literature on BWS - filename: figure1.eps



Statistical analysis

The selected studies (Table 2) presented the investigation of gait characteristics at different values of the percentage of BWU, ranging from 0% to 100%. These characteristics were identified by surveying the literature listed in Table 1 and then grouped together based on the nature of these parameters (Table 3). However, these values were not uniform across the studies and were usually in increments of 10% to 20% BWU. Linear interpolation was used to obtain the values of gait parameter results at every 5% of unloading. This allowed comparison between studies at all percentages of BWU and bolstered the analysis by providing more data. However, no extrapolation was applied for the parameter results. Each parameter was normalized by taking a ratio with respect to its value at 0% BWU in a given study. This way the scaling process brought an uniformity in results and allowed comparison of trends across literature. By removing the dimensions attached to each parameter through scaling, comparison across different gait parameters was possible. For each of the four categories mentioned above, the mean and standard deviation (SD) for all gait parameters was calculated using the results of the relevant studies.

Linear regression was used to further analyze the response of the gait characteristics to the increase in % BWU. The slope (m) and the coefficient of determination (R^2) for the gait parameters are presented in the results section. The slope 'm', which represents the change in the normalized parameter value per unit change in

Table 2 Studies considered in statistical analysis

	Treadmill-based	Overground
Modulated BWU	Franz 2007, Franz 2008 – *2 studies*	Fenuta 2014 - 1 study
Constant BWU	Finch 1991, **Dietz 1997**, **Dietz 1998**, Colby 1999, **Hesse 1999**, Stephens 1999, Griffin 1999, **Danielsson 2000**, Ferris 2001, Ivanenko 2002, Threlkeld 2003, Grabowski 2004, Van Hedel 2006, **Phadke 2007**, Thomas 2007, McGowan 2008, Aaslund 2008, McGowan 2009, Lewek 2010, Goldberg 2013, Van Kammen 2014, Worthen-Chaudhari 2015, Dragunas 2016 – *23 studies*	Patino 2007, **Sousa 2009**, **Burgess 2010**, Wang 2011, Serrao 2012, Barela 2014, Fischer 2015, Fischer 2015*, **Hurt 2015**, **Jung 2016**, Fischer 2016, Mun 2016 – *12 studies*

the % BWU, has units $\%^{-1}$. An 'm' value less than or equal to 1 x 10^{-3} $\%^{-1}$ was approximated as zero and the parameter was assumed to remain constant. Negative R^2 values were reported as 0. They indicate that, in a given category, the % BWU was not a useful predictor for that gait parameter. A R^2 value above 50% was considered as a good fit. For a given category (TH, OH, etc.), the R^2 value was only calculated if the number of available raw data points was higher than 3. Since the data was normalized, for each category, the zero conditions for all the relevant studies were considered as one data point in total.

Results

Study of gait characteristics

This section details the parameters used to study and characterize human gait in general but especially for the experiments concerning BWS systems. For each of these parameters, the results from the studies were plotted against the % of BWU support, from respective studies. The aim of these plots is to understand if the parameters follow a specific pattern with respect to the % BWU. A summary of the results for all gait parameters is presented in Table 5.

Table 3 Categorization of gait characteristics

Group	Parameters
Gait spatio-temporal parameters	1. Stride length 2. walking speed 3.cadence 4. single limb support phase 5. double limb support phase 6. total stance phase
Joint kinematics	7. Hip angle range of motion (ROM) 8. knee angle ROM 9. ankle angle ROM
Joint kinetics	10. Hip extension moment 11. hip flexion moment 12. knee extension moment 13. knee flexion moment 14. ankle joint moment 15. ankle joint impulse
Ground reaction forces (GRF)	16. Antero-posterior peak I 17. Antero-posterior peak II 18. vertical GRF peak I 19. vertical GRF peak II
Energy consumption	20. Energy cost of walking (ECW) 21. Heart rate (HR)
Electromyography data (EMG)	Muscle activity for muscles - 22. MG 23. LG 24. RF 25. BF 26. TA

Apart from the gait characteristics presented in Table 3, other characteristics have been investigated in the literature concerning BWS systems, such as: (1) gait symmetry [12, 77]; (2) consistency of gait cycles [58]; (3) trunk movement [12, 27, 61]; (4) pelvic motion [61, 82]; (5) leg segment kinematics [12]; (6) joint power generation [31, 56, 83]; (7) nocioceptive flexion reflex [84]; (8) soleus H-reflex [25, 85]; (9) vertical impulse [86] and (10) horizontal trunk work [86]. These gait parameters were not analyzed either because there was only one study about them or in case of multiple studies, the available data was in a form that did not allow for comparison across literature.

Table 4 Selected conditions for statistical analysis [13, 63, 83, 87–92]

Study	Condition
Stephens et al. 1999	0.9 - 1 ms^{-1}
Ivanenko et al. 2002	1.1 ms^{-1}
Van Hedel et al. 2006	1.5 ms^{-1} (2 ms^{-1} for joint angles)
Thomas et al. 2007	1.26 ms^{-1}
Aaslund et al. 2008	1.2 ms^{-1}
Lewek et al. 2010	1.2 ms^{-1}
Van Kammen et al. 2014	1.8 ms^{-1}
Dragunas et al. 2016	1.47 ms^{-1}

In case of the studies with treadmill, especially for healthy subjects, some of them investigate the gait characteristics at multiple walking speeds in addition to different BWU levels. In order to analyze their results together, the outcomes for a specific walking speed are selected. The experiment by Threlkeld et al. was conducted only at a single treadmill speed of 1.25 ms^{-1} [93]. In order to compare the results from this study, data from other treadmill-based experiments with multiple speed conditions was chosen at the speeds shown in Table 4.

For the literature comparing modulated and unmodulated BWU, only the results for the unmodulated condition were used for the statistical analysis [31, 77]. This enabled the comparison with other papers which use only unmodulated BWU. The literature on modulated BWU investigated the gait characteristics at 0% and 20% BWU [31, 77]. The difference in outcomes for modulated and unmodulated conditions as compared to the 0% BWU condition is presented separately.

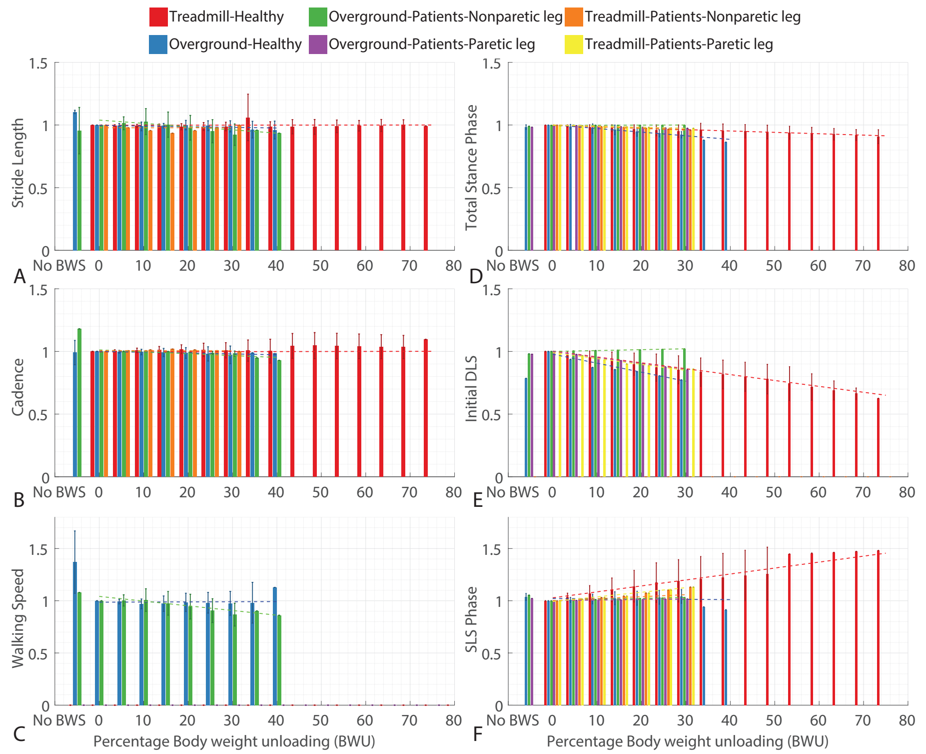
Gait spatio-temporal parameters

The values (Figure 2A) of stride length (SL) were reported by 12 studies [8, 12, 13, 28, 31, 83, 89, 90, 93–96]. None of four groups showed a specific behavior for the magnitude of stride length. In case of the experiment by Franz et al., SL changed by -3% for the unmodulated 20% BWU as compared to -1% for the 20% modulated BWU [31]. 10 papers described the influence of % BWU on cadence (Figure 2B) [8, 12, 27, 28, 31, 89, 90, 93–95]. OP group presented a decrease in cadence while other three groups did not present any definite pattern. Modulated BWU at 20% support led to a -0.78% difference in cadence in comparison to -3.2% difference for unmodulated BWU [31].

Data for walking speed (Figure 2C) was extracted from 5 studies [8, 12, 28, 95, 96]. The OH groups showed a considerable decrease in gait speed from walking without harness to 0% BWU but no specific behavior beyond 0% BWU. In case of OPN group, gait speed did not display any particular trend for BWU greater than 0%. The results from the experiments involving treadmill were not presented since the participants usually walk at a predetermined speed on the treadmill.

Results for the proportion (percentage) of total stance phase (ST) (Figure 2D), initial double-limb support phase (iDLS) (Figure 2E) and single-limb support phase (SLS) (Figure 2F) in the gait cycle were taken from 10 studies [8, 12, 28, 87–89, 92–94, 96]. The proportion of swing phase (SW) and terminal double-limb support phase (tDLS) can be inferred from the above presented values. ST remained almost constant for all groups except TPN and OH, where it decreased. ST also decreased for the TH group but there was no agreement within the studies for the slope (m) of the decrease. iDLS stayed constant for the OPN category but reduced in case

Figure 2 Gait spatio-temporal parameters where DLS: Double limb support, SLS: Single limb support Results for the gait spatio-temporal parameters - filename: figure2.eps



of all other groups. SLS did not show a specific pattern for the OH group while it remained unchanged for the OPP group. However, SLS increased for other four groups.

It is important to note that data for gait phases for the OPN and OPP groups was obtained from a single research paper [12]. Furthermore, results for all the spatio-temporal parameters for the TP category were also available from only study [94].

Joint kinematics

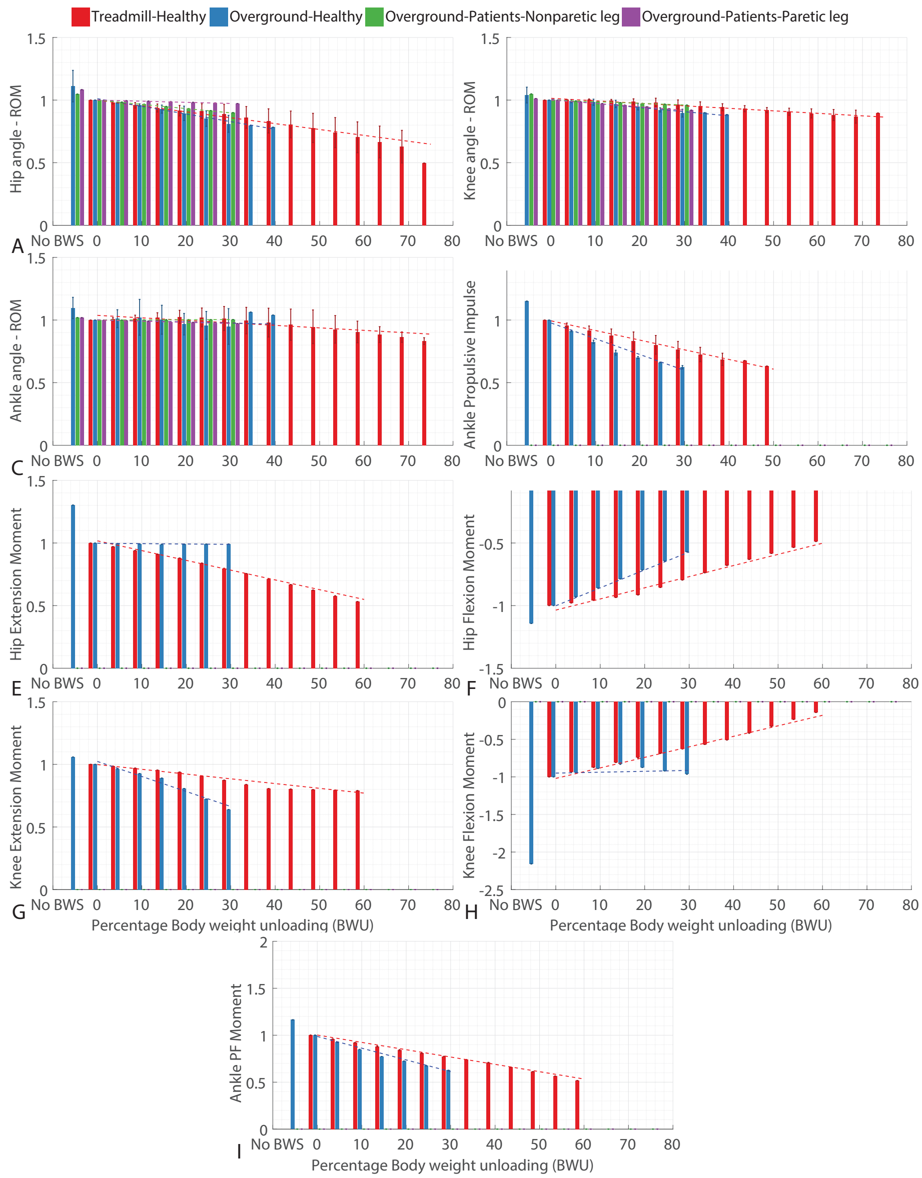
The ROM (range of motion) data of all three leg joints for the overground-patient group (OPN and OPP) was obtained from a single study [12]. The R^2 values for these two groups are 100% as this study contained data for only two conditions, 0% and 30% BWU [12].

Data for hip joint ROM (Figure 3A) was analyzed from 7 studies [8, 12, 28, 31, 89, 93, 96]. Hip ROM decreased for TP, OH and OPN groups but remained roughly unchanged for the OPP group. However, in case of the OH group, the ROM reduced considerably after 30% BWU. The change in hip ROM for modulated and unmodulated BWU at 20% was -1.21% and -11.41% respectively [31].

6 studies were used to obtain the data on knee joint ROM (Figure 3B) [8, 12, 28, 89, 93, 96]. Rise in % BWU led to a reduction in knee joint ROM for all four groups. Ankle joint ROM (Figure 3C) results were extracted from 8 studies [8, 12, 28, 31, 85, 89, 93, 96]. Ankle ROM almost remained constant for the neurologically impaired participant groups i.e. OPN and OPP. Contrary to this, it did not show

any specific behavior for the healthy groups i.e. TH and OH. In case of modulated BWU, modulating led to 5.86% change in ankle ROM as compared to the 5.21% for unmodulated unloading [31].

Figure 3 Joint kinematics and joint dynamics Results for the joint parameters - filename: figure3.eps



Joint dynamics

The data for hip and knee moments was obtained from 2 studies [8, 97], for ankle plantarflexion (PF) moment from 3 studies [8, 83, 97] and ankle propulsive impulse from 4 studies [8, 83, 98, 99]. Except for ankle propulsive impulse, data for the OH [8] and TH [97] groups for all other parameters was obtained only from one study each. In Figure 3, flexion and extension moments are presented with negative and positive signs respectively to indicate opposite directions.

Ankle impulse and ankle PF moment decreased for both the TH and the OH groups. Hip extension moment and knee flexion moment remained roughly constant for the OH group (up to 30% BWU) while they reduced for the TH group. However, hip flexion and knee extension moments reduced for both the groups.

Ground reaction forces

Data for the anteroposterior (AP) ground reaction force (GRF) was obtained from 5 papers [28, 31, 83, 98, 99] and for the vertical GRF from 6 papers [28, 31, 83, 88, 94, 98]. However, it should be noted that the data for the GRF for the TP group was from a single research study [94].

The peak values of AP GRF (AP GRF I - negative and AP GRF II - positive peaks) and vertical GRF (first and second peak) were considered for the statistical analysis. AP GRF values decreased for both TH and OH categories while vertical GRF reduced in magnitude for the TH, TP and OH categories. The reduction was consistently larger for the OH group as compared to the other two groups.

For the AP GRF I and 1^{st} vertical GRF peaks, the results for 20% modulated unloading were closer to 0% BWU for TH group than the 20% unmodulated unloading [31, 31]. However, for the AP GRF II and 2^{nd} vertical GRF peaks, it was vice-versa [31, 77].

Figure 4 Ground reaction forces and energy consumption where GRF: ground reaction forces
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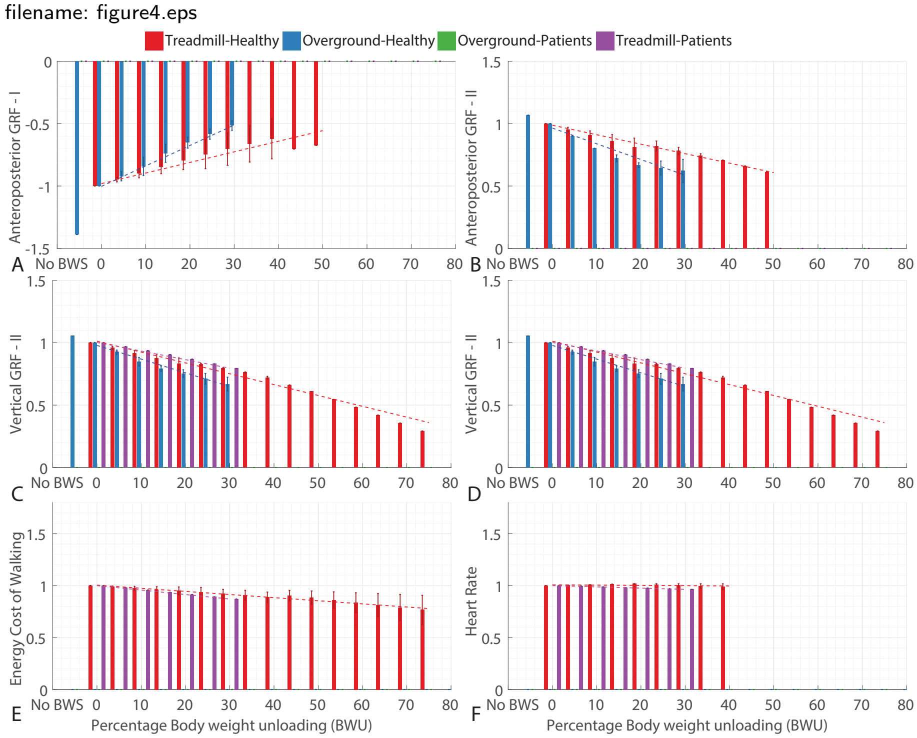


Energy consumption and heart rate

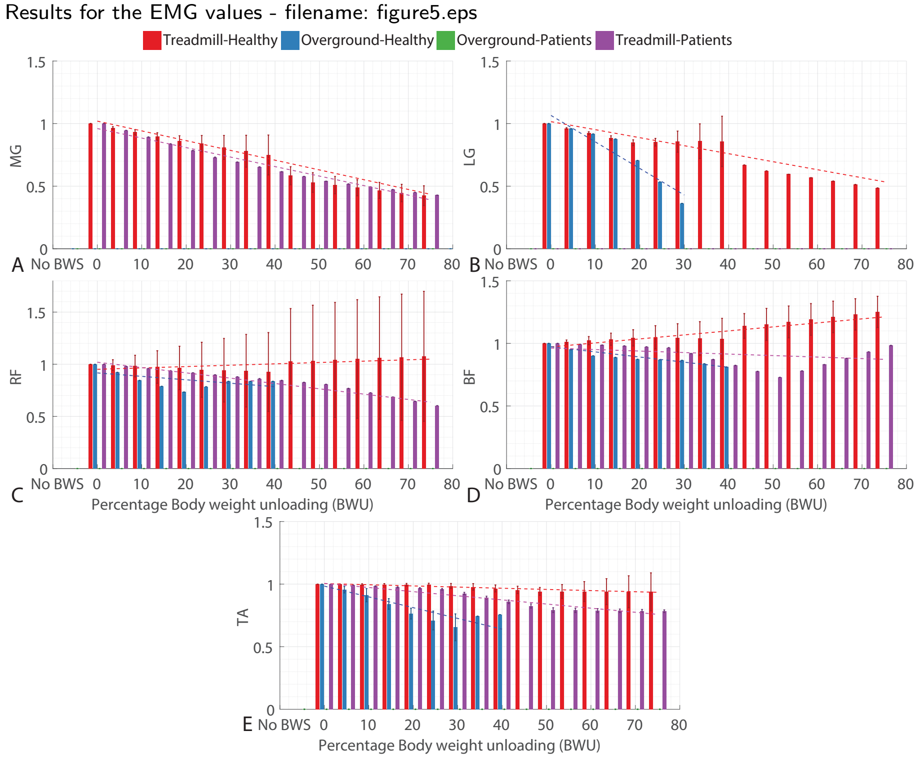
Outcomes for energy cost of walking (ECW) were acquired from 5 studies and reported in terms of the VO_2 max (volume of maximal oxygen uptake) [90, 100–103]. ECW (Figure 4E) showed a similar decreasing trend for both the TP and the

TH groups. Data for heart rate was obtained from 3 papers [90, 101, 102]. Heart rate (Figure 4F) did not show any specific trend for the TH category while it showed slight reduction for the TP category.

Muscle activity

Muscle activity was considered in terms of the magnitude of the EMG signal as an average value over the entire gait cycle. Five muscles were examined: (1) medial gastrocnemius (MG); (2) lateral gastrocnemius (LG); (3) tibialis anterior (TA); (4) rectus femoris (RF) and (5) biceps femoris long head (BF). Apart from the studies considered for statistical analysis (mentioned below), other studies also investigated the influence of BWU on muscle activity [28, 54, 81, 87, 89, 92, 94, 99, 104, 105]. However, the relevant data for the average value of EMG signal from these papers was not available and hence they are excluded.

Figure 5 Mean muscle activity over gait cycle where MG: medial gastrocnemius, LG: lateral gastrocnemius, TA: tibialis anterior, RF: rectus femoris and BF: biceps femoris long head
Results for the EMG values - filename: figure5.eps



Extensor muscles

MG muscle (Figure 5A) showed a decrease in muscle activity with the increase in % BWU for both the TH [83, 85, 101, 106, 107] and the TP categories [106, 107]. LG muscle (Figure 5B) presented a reduction in magnitude for both the TH [83, 88] and the OH groups [108]. For the RF muscle (knee extensor, Figure 5C), two groups, TH [88, 101, 106] and OH [96] did not show a any clear trend while the TP group [106] presented a decrease in the magnitude of muscle activity.

Flexor muscles
In case of the BF muscle (Figure 5D), the TP group [106] and TH group [88, 101, 106] failed to show any consistent pattern in muscle activity while the OH group showed a clear decrease [96]. TA muscle (Figure 5E) activity reduced for the TP [106, 107] and OH [96, 108] groups but did not present a consistent behavior for the TH group [85, 88, 106, 107].

Summary

The above presented results are summarized in Table 5.

Gait Characteristics	Treadmill						Overground					
	Healthy			Patients			Healthy			Patients		
	n	m	R^2	n	m	R^2	n	m	R^2	n	m	R^2
		$\times10^{-3}$	%		$\times10^{-3}$	%		$\times10^{-3}$	%		$\times10^{-3}$	%
1. Stride length	5	*-0.03**	0	1	*0**	0	4	*-0.5**	4.1	2	*-2.6**	29.7
2. Cadence	5	*0.1**	0.1	1	*0**	0	3	*-0.7**	5.4	2	-1.5	65.1
3. Walking speed		—			—		4	*0.1**	0	2	*-4.5**	48.5
4. Gait phases - ST	4	*-1.1**	41.7	1	-1.4; **-0.9**	NA	3	-2.9	71.3	1	**0.1**; **-0.8**	NA
5. Gait phases - iDLS	3	-4.7	74.8	1	—; -4.7	NA	1	-7.2	93.6	1	**0.7**; -4.9	NA
6. Gait phases - SLS	2	5.7	61.9	1	1.9; 4.4	NA	3	*-0.3**	0.5	1	1.1; **0.6**	NA
7. Hip joint ROM	3	-4.7	76.6		—		3	-6	80.3	1	-3.4; **-0.9**	NA
8. Knee joint ROM	2	-2	80.8		—		3	-3.3	79.3	1	-1.3; -2.7	NA
9. Ankle joint ROM	4	*-2**	37.8		—		3	*-0.9**	1.4	1	**0.1**; **-0.9**	NA
10. Ankle impulse	2	-7.7	93.9		—		2	-12.6	94.9		—	
11. Hip E moment	1	-7.8	99.2		—		1	**-0.3**	NA		—	
12. Hip F moment	1	-8.9	97.1		—		1	-14	NA		—	
13. Knee E moment	1	-3.8	92.8		—		1	-12	NA		—	
14. Knee F moment	1	-14	98.9		—		1	**-1**	NA		—	
15. Ankle PF moment	2	-7.8	99.4		—		1	-12	NA		—	
16. AP GRF peak - I	3	-8.5	80.2		—		2	-16.4	96.3		—	
17. AP GRF peak - II	3	-7.6	91.6		—		2	-12.6	87.7		—	
18. Vertical GRF - I	3	-8.3	95	1	-6.6	NA	2	-9.6	99.1		—	
19. Vertical GRF - II	3	-8.7	96	1	-6.9	NA	2	-11	93.8		—	
20. ECW	5	-3	70.2	1	-4.3	NA		—			—	
21. Heart rate	2	*-0.3**	6.6	1	-1.2	NA		—			—	
22. EMG - MG	5	-7.8	83.1	2	-7.6	96.7		—			—	
23. EMG - LG	2	-6.4	72.1		—		1	-21.2	NA		—	
24. EMG - RF	3	*1.3**	1.3	1	-5.1	95.9	1	*-3.3**	30.5		—	
25. EMG - BF	3	*-3.2**	42.5	1	*-1.2**	8.4	1	-4.1	88.6		—	
26. EMG - TA	4	*-0.9**	12.4	2	-3.3	88	2	-8.76	73.3		—	

Table 5 Summary of the change in gait characteristics influenced by the BWU level, where *italic**: no definitive trend across studies, **bold:** gait parameter remains almost constant, —: no studies. Number of studies (n), slope (m %$^{-1}$) and % adjusted R^2 values for the linear regression of each gait parameter are written respectively. In cases where the magnitude of gait parameters is measured separately for non-paretic (T/OPN) and paretic legs (T/OPP), slope for the non-paretic leg is mentioned first. E: extension, F: flexion, ECW: energy cost of walking

Discussion

This paper combined all studies on the effect of BWU on the gait in order to analyze how body weight unloading influences gait characteristics. There are four topics that still have to be addressed after reviewing the results: (1) the general trends in how BWU influences different gait parameters (2) the difference between the influence of BWU in treadmill and overground walking (3) a comparison between modulated and unmodulated BWS and (4) an overview of the literature on BWS studies.

On the amount of BWU

The optimum amount of BWU is an important factor for gait rehabilitation training and consequently a key topic of studies on the effects of BWU on gait [24]. From the results of this paper, it can be seen that the increase in the amount of BWU influenced all the 26 gait parameters listed in Table 5. While the percentage of single limb stance (SLS) phase increased with the increase in BWU, almost all other parameters showed a decreasing trend. Three parameters (Stance phase, Heart rate & TA muscle activity) showed inconsistent behavior for only TH group, two (Ankle joint ROM & RF muscle activity) for both TH and OH groups, BF muscle activity for TH and TP groups, and walking speed for OH and OP groups (Table 5). Stride length and walking speed did not present a consistent behavior for all relevant groups (Table 5).

Considering the patient group walking overground, the gait spatio-temporal parameters like cadence and gait phase proportions, and the kinematic parameters like ankle and knee ROM show a weak influence of unloading force up to 30% BWU. However, 13 studies (9 out of 16 for overground walking) investigated the effects of %BWU only up to 30%. For gait characteristics and participant groups where the R^2 values lies between 50% and 60%, there is usually a similar trend (downward/upward) for all considered studies but little consistency in the slope (m) values across these studies.

In case of healthy participants, for both the treadmill and overground walking, the relative magnitude of change in joint dynamics, GRF, energy cost of walking and muscle activity is higher than that in gait kinematics and spatio-temporal parameters (Table 5). The implications for the decrease in energy cost of walking are discussed in detail later in this section.

Curvature patterns of the joint trajectories remain roughly similar despite of the increase in BWU level up to 30% [8, 12, 28, 31, 58, 58, 89, 93, 96]. It is possible that the changes in the hip and knee adduction moment and ankle propulsive impulse and the changes in COP trajectory allow the kinematic patterns to remain similar [88, 109]. Thus, it can be inferred that up to 30% BWU force can be applied without significantly modifying the kinematic and spatio-temporal parameters associated with gait, which may be beneficial for the outcome of treatment [30]. This result from the meta-analysis aligns well with what other researchers already suspected in their separate studies [8, 54, 56, 76, 89, 93]. Of course, in some cases a higher amount of BWU might be necessary, for instance when patients find it difficult to bear their weight even with 30% BWU.

Comparison with literature on reduced gravity

A comparison with the conclusions of the paper by Richter et al. and the research by Harvill et al. is presented here [46, 47]. Harvill et al. studied the effects of reduced gravity on gait for the purpose of space exploration while the paper by Richter et al. reviewed other literature on this topic. Regarding gait spatio-temporal parameters, both the papers reported a decrease in stance phase duration, a corresponding increase in swing phase duration but no specific trend for stride length and cadence. Richter et al. noted a higher dependence on walking speed for both stride length and cadence. These results are in agreement with our findings (Table 5).

In case of joint kinematics, both of these papers described a reduction in hip ROM and knee ROM. Harvill et al. noted a decrease in ankle ROM contrary to the inconsistent behavior reported by Richter et al. However, Richter et al. noted a very high effect size for hip and knee ROM unlike our results which show a weaker influence (Table 5). A possible explanation for this difference is that Richter et al. only analyzed gait parameters at very high (> 60%) BWU levels.

According to Richter et al., joint impulses, energy cost of walking and heart rate showed higher reduction compared to kinematic parameters due to the decrease in gravity. GRF presented the highest influence of gravity in both the studies. In addition to showing that joint moments also show a large influence of simulated gravity (BWU level), our findings corroborate these results. The only exception is heart rate, for which we did not find any consistent behavior. Joint moments, impulses and GRF directly reflect the oscillation of the COM during gait, so their changed behavior under BWU shows that gravity plays an important role in COM oscillation.

Energy cost of walking

Table 5 shows that energy cost of walking decreases with the increase in BWU level for the TH group. Studies by Richter et al. and Harvill et al. report a similar trend. [46, 47]. An earlier review by Wortz et al. also states that at lunar gravity (similar to around 83% BWU), human locomotion entails significantly lower energy cost than at terrestrial gravity conditions (similar to 0% BWU) [110]. However, this reduction in energy requirement is not limited to a walking gait. In fact, as the BWU level is raised or the effective gravity lowered, the energy cost for a running or skipping gait decreases more rapidly than the cost for walking gait [68, 111]. Thus, at high BWU levels, walking is not the cheapest mode of locomotion in terms of energy cost. It is hypothesized that leg movement and thus the mode of locomotion is modulated to minimize the energy consumption during locomotion [112, 113]. This might lead to changes in gait at high levels of unloading which would be difficult to detect due to the smooth transitions [69, 114], thus adversely affecting the task specificity of BWS training.

On treadmill vs overground studies

Comparison of results for the gait in overground (OG) and treadmill (TM) studies shows small but important differences (Tables 5 and 6) in gait characteristics. The OH group presents a greater influence of BWU on all gait parameters except single limb stance phase, hip extension moment and knee flexion moment. The TH group shows greater influence for these three parameters. Only in case of gait phases, neurologically impaired individuals show relatively similar results for both the walking conditions. This is in agreement with the conclusions from existing literature on walking without body weight support. If the treadmill speed is not set to match the preferred overground walking speed, differences arise between treadmill and overground walking [115–119]. These differences are prominent if the participants walk at self-selected walking speed on the treadmill which is not equal to the preferred speed in overground walking [120]. Thus, if the participants are not able to attain the preferred overground speed on a treadmill, the training might lose its task-specific nature [121].

Walking on a treadmill shows that both the treadmill speed and the amount of unloading have considerable influence on gait parameters [83, 89, 92, 97, 98, 122–124]. Cadence and stride length are affected more by the treadmill speed than by

Table 6 Summary of data in Table 5 – Trends for gait parameters which show different behavior in TM and OG environments

Affected parameter	Group	Treadmill	Overground
1. Cadence	Patients	Inconsistent	Decreasing
2. Stance phase %	Healthy	Inconsistent	Decreasing
3. SLS phase %	Healthy	Increasing	Inconsistent
4. BF muscle activity	Healthy	Inconsistent	Decreasing
5. TA muscle activity	Healthy	Inconsistent	Decreasing

the percentage of BWU, except for above 75% BWU [46, 89]. The relative duration of gait cycle phases and consequently the joint angle patterns and the muscle activation patterns are influenced by the treadmill speed. Joint torques, ankle power generation, GRF profiles and pelvic excursions are also affected.

Treadmill walking may lead to confounding effects of BWU on gait characteristics and reduce the effectiveness of BWST [109]. While the data from the overground experiments shows that the walking speed changes beyond 10 % of unloading, treadmill forces the participants to walk at a constant speed, which can result in unnatural gait dynamics. However, modulation of treadmill speed according to the amount of unloading provided might help to retain the natural gait pattern.

In case of the OH group, there was a reduction in gait speed from unsupported locomotion to walking in a harness at 0% BWU. A reasonable explanation for this observation is the requirement from users to pull the BWS system along while walking. Though overground walking seems more suited to gait training, pulling the BWS system forward against resistance, for example caused by friction, can make it difficult for the users to maintain a comfortable walking speed [12, 28]. However, using a motor-actuated winch system to pull the BWS system may help to ameliorate this problem [8].

On modulated vs unmodulated support.

In the method section, we made a distinction between modulated and unmodulated support. Although there has been little research into modulated support, this section discusses the potential benefits of such systems as found in literature.

A BWS system should account for an individual's specific physiological limitations and promote efficient locomotion patterns in order to provide optimal rehabilitation [125]. It has been claimed that modulation of unloading force can enable appropriate ground contact and limb motion while allowing gait spatio-temporal parameters like walking speed, cadence and stride length to be comparable to the values during unsupported walking [31]. Franz et al. designed a BWS system which controlled the unloading force based on gait cycle phases and conducted an experiment to compare it against a BWS system with constant unloading force [31, 77]. They compared the difference in the values of stride length, cadence, hip and ankle joint ROM, ankle power generation and GRF for constant and modulated 20% BWU conditions. The movement patterns and the magnitude of these parameters, except for anteroposterior GRF (deceleration) and 2$^{\text{nd}}$ peak of vertical GRF, were closer to unsupported walking in case of modulated BWU.

Van Thuc et al. followed another approach towards modulation of unloading force; controlling the direction and magnitude of force according to the center of pressure

(COP) trajectory [78]. They observed that the COP trajectory produced as a result of modulated BWU resembled that of unsupported walking more closely, as compared to the one with constant BWU.

Munawar et al. controlled the unloading force with the aim of offsetting the inertial forces of the user's body dynamically [79]. They reported the pattern of vertical GRF for modulated BWU to be similar to that of unsupported walking. Ivanenko et al. and Fenuta et al. also conducted experiments with a modulated BWS system but did not report any comparative results between modulated BWU and constant BWU [81, 88]. Thus, it can be concluded prima facie that modulated unloading force generally led to less difference from the 0% BWU condition than unmodulated unloading.

On body weight support studies

This paper compared 55 studies in the terms of the effects of the BWU on gait, published from the year 1991 to 2016 (refer to `additional file 1`). Of these 55 studies, 30 are from the period 1991–2010 (20 years) while the remaining 25 are published 2010 onwards (6 years). This shows an increasing interest in the potential of BWST as a neurological rehabilitation tool.

Only 27% of the studies are based on individuals with either one of the neurological disorders (Stroke, Spinal cord injury and Parkinson's disease) as participants (Figure 1). This proportion is low considering that neurologically impaired individuals are the primary user group for BWS systems in general and rehabilitation tools in particular. The number of participants for the studies ranges from 1 to 28, with an average of 12 participants. In addition to this, only 53% of these publications explicitly state the use of randomization in the experimental protocol. This is in stark contrast to clinical studies, which generally include higher number of participants and are randomized by design [126–128]. Clinical studies were not included in this review since we could not find clinical trials which also presented gait parameter data during BWS training along with the post-training data. Secondary outcomes presented by clinical studies are also generally only assessed after training and so without BWS. The review by Richter et al. reported a similarly low methodological quality of studies investigating the influence of low gravity on human gait [46]. The low number of participants and the lack of randomized trials can both lead to suboptimal study design [129]. The proportion of studies investigating modulated BWS systems is around 13%, with the average number of participants being 6. These low numbers indicate a strong necessity for further research on BWS systems providing modulated unloading force.

The amount of BWU used for experiments ranges from 5% to 100%, with almost all studies utilizing different combinations and magnitudes of BWU (Figure 1). Apart from the amount of unloading, the gait characteristics tested also vary substantially from one study to another. This suggests that there is no agreement within the research community over the appropriate levels of BWU for testing and the relative importance of gait parameters to be examined.

Limitations of this review

The limitations of this study are presented and discussed below. First and foremost limitation is that the results from different patients groups (SCI, stroke and

Parkinson's disease) are pooled together and analyzed as a whole. This was done due to the paucity of studies based on subjects with a neuromuscular disorder. Pooling results together provided a large enough sample size for a meaningful statistical analysis. In order to minimize the distortions in the results due to different pathologies, data from each paper was normalized with respect to the value at 0% BWU. This normalization process shifted the focus of the analysis from absolute values of gait parameters to the trends followed by these parameters.

The second limitation is the combined analysis of experimental results based on different BWS systems. There are not enough studies for each BWS system to analyze the results separately. The third limitation is that experiments differing in usage of arms were also pooled together due to the limited number of studies. However, to improve consistency of data, only vertical BWS systems based on a harness-based attachment system were included in the analysis. This decision was taken based on the assumption that evaluating one only type of BWS system will reduce the artifacts introduced in the results by the BWS system.

Finally, this review is limited by the lack of a single metric to characterize and compare human gait. Furthermore, it is difficult to rank gait characteristics based on their importance to gait. Depending on the context, a small change in one gait parameter might be more important than a larger change in another. As a result of this, gait parameters were selected based on their frequency of use in practice, and a large number of gait parameters (26 in total) were analyzed, despite the scarcity of relevant studies for some of these parameters.

Conclusion

This paper studied the influence of body weight unloading (BWU) on gait parameters through a meta-analysis. The results were grouped based on the physiological condition of the subjects (healthy or neurologically impaired) and on the type of walking environment (treadmill or overground). The BWS systems were categorized based on the nature of unloading force (un/modulated) and the type of walking environment. It was observed that dynamic characteristics of gait were more influenced by BWU than the kinematic ones but there is no consensus in literature for some of these parameters. However, upto 30% unloading, the influence of BWU on kinematic gait parameters seemed to be limited. This finding has wider implications on the effectiveness of BWST, since a natural gait may be maintained below 30% unloading. The distinction and subsequent investigation of these gait characteristics may help to unearth pivotal compensatory mechanisms in gait and serve as a reference document for conducting future studies on the effects of body weight unloading on human gait.

List of abbreviations

BWS: Body weight support
BWU: Body weight unloading
BWST: Body weight supported training
SCI: Spinal cord injury
BWSTT: Body weight supported treadmill training
TH: Treadmill-healthy
OH: Overground-healthy
TP: Treadmill-patients
OP: Overground-patients

TPN: Treadmill-patients non-paretic leg
OPN: Overground-patients non-paretic leg
TPP: Treadmill-patients paretic leg
OPP: Overground-patients paretic leg
T-M: Treadmill modulated
O-M: Overground modulated
T-UM: Treadmill unmodulated
O-UM: Overground unmodulated
ROM: Range of motion
GRF: Ground reaction forces
AP: Antero-posterior
ECW: Energy cost of walking
HR: Heart rate
EMG: Electromyography
ST: Stance phase
iDLS: Initial double limb stance phase
SLS: Single limb stance phase
SW: Swing phase
tDLS: Terminal double limb stance phase
MG: Medial gastrocnemius
LG: Lateral gastrocnemius
TA: Tibialis anterior
RF: Rectus femoris
BF: Biceps femoris long head
COP: Centre of pressure

Declarations
Not applicable

Consent for publication
Not applicable

Availability of data and material
Please contact author for data requests.

Funding
Horizon 2020 Framework Programme, Award Number: 10 152, Recipient: Heike Vallery

Competing interests
Heike Vallery and Michiel Plooij have been and continue to be involved in the design and commercialization of body-weight support systems.

Author's contributions
SA collected, analyzed and interpreted the data regarding the gait characteristics and was the main contributor to the manuscript. MP and HV checked scrutinized data, the analysis process and the results. In addition, MP and HV supervised the writing process for the manuscript. All authors read and approved the final manuscript

Acknowledgements
This work is (partially) funded by Eurostars Project 10 152 – LEAP.

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4

GAIT MODELS

4.1. SIMPLEST WALKING MODEL

This section presents a basic description of the Simplest walking (SW) model by [40], followed by the derivation of the equations of motion for the three BWS strategies. Assuming flat surface ($\gamma = 0$), the equations of motion from [40] were modified to model the three BWS strategies.

MODEL DESCRIPTION

The Simplest walking (SW) model has two rigid massless legs with infinitesimal point-masses at the feel. These legs are hinged to the hip, which also has a point-mass (Figure 4.1) . This model is a simplified version of the walker by McGeer [41]. The linked mechanism moves on a ramp with slope γ. When a foot hits the ramp surface during heel-strike, it has a plastic collision (no-slip, no-bounce) and the velocity of the foot becomes zero. The swing foot is assumed to pass through the ramp surface and the foot scuffing phenomenon (geometric interference) is ignored for the sake of simplicity. When the swing foot hits the ramp at heelstrike, there is an impulse at the contact point. However, it is assumed that the former stance leg has no impulsive reaction with the ramp when the leg leaves. Based on this assumption, angular momentum is conserved through the collision for the new swing leg about the hip and the whole body about the swing foot contact point. For more details about the model, please refer to [40].

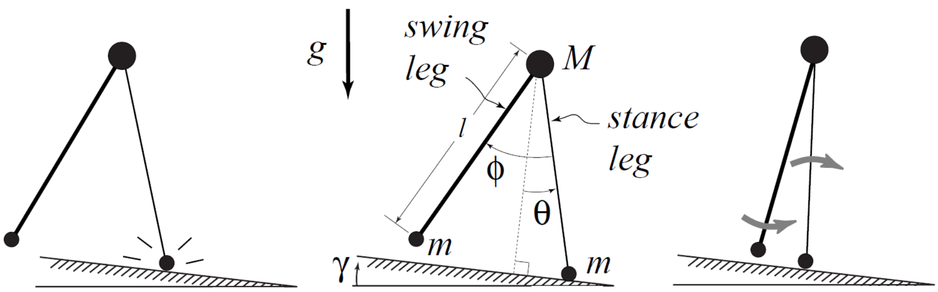


Figure 4.1: Schematic of the Simplest walking model. Figure reprinted from [40]

Model parameters for the SW model are:

- 'm' is the mass of the foot
- 'M' is the mass at hip
- (x_1, y_1) is the location of the hip mass
- (x_2, y_2) is the location of the swing foot mass
- Stance foot is at the origin

- $\theta(t)$ represents the stance leg angle w.r.t. to the slope normal
- $\phi(t)$ is the angle between the swing and the stance legs
- γ is the ramp slope
- β is the percentage of BWU provided and u is equal to $\beta/100$

Position and velocity of the hip mass (x_1, y_1) and swing foot (x_2, y_2) mass are given by:

$$
\begin{aligned}
x_1 &= -l\sin\theta & \& \quad & x_2 = -l\sin\theta - l\sin(\theta-\phi) \\
\dot{x}_1 &= -l\cos\dot{\theta}\theta & \& \quad & \dot{x}_2 = -l\dot{\theta}\cos\theta - l(\dot{\theta}-\dot{\phi})\cos(\theta-\phi) \\
y_1 &= l\cos\theta & \& \quad & y_2 = l\cos\theta - l\cos(\theta-\phi) \\
\dot{y}_1 &= -l\dot{\theta}\sin\theta & \& \quad & \dot{y}_2 = -l\dot{\theta}\sin\theta + l(\dot{\theta}-\dot{\phi})\sin(\theta-\phi)
\end{aligned}
$$

'Foot' is assumed to be much smaller than the 'body' and so:

$$\frac{m}{M} = 0$$

The transition rule at heelstrike collision is given by:

$$\phi(t) - 2\theta(t) = 0$$

Actuation principle

For the simulations in this work, the model is assumed to walk on a flat surface ($\gamma = 0$). Thus, in order to compensate for the energy loss during heelstrike, the actuation principle from [42] was used. This entails the application of a toe-off impulse, acting on the trailing foot just before heelstrike. This impulse is directed towards the centre of mass (hip). This impulse emulates the ankle plantarflexion and knee extension which occurs just before the toe-off motion. This impulse is given by –

$$P = \phi\tan\alpha$$

where ϕ is the angle of the swing leg w.r.t to the stance leg just before heelstrike and α is the angle of the stance leg w.r.t to the vertical at the time of heelstrike. The impulse term P modifies the transition rule at heelstrike collision. The second method of actuation is a spring-like hip torque applied to the swing leg. This torque emulates the burst in hip extensor activity at the end of swing phase and hip flexor activity immediately after toe-off. Due to the negligible mass of the swing foot, this torque does not influence the motion of the stance leg nor the overall mechanical energy. The torsional spring constant (k_f) for this spring is dimensionless and normalized by leg length and body weight. The hip torque acts for the entire swing phase and is given by:

$$\text{Hip torque:} \quad \tau_{\text{hip}} = -k_{\text{f}}\phi \tag{4.1}$$

Constant force

A term representing the constant vertical unloading force ($F_u = uMg$) can be added to the original equations [40]. Alternately, the term for gravity g can be replaced by the term $(1-u)g$ for simulated reduced gravity. The second approach was used here and time is scaled by a factor of $\sqrt{\frac{l}{(1-u)g}}$. Furthermore, considering the hip torque, the resulting equations of motion are:

$$\ddot{\theta} = (1-u)\sin\theta \tag{4.2}$$

$$\ddot{\phi} = \ddot{\theta} + \dot{\theta}^2\sin\phi - (1-u)\cos\theta\sin\phi - k_{\text{f}}\phi \tag{4.3}$$

Counterweight system

Mass of the counterweight is 'uM' where where 'u' represents the amount of BWU provided as a ratio of body mass and M is the mass at the hip.

Potential energy 'V_{cw}' of the system:

$$V_{\text{cw}} = (M - Mu + m)gl\cos\theta - mgl\cos(\theta-\phi) \tag{4.4}$$

Kinetic energy 'T_{cw}' of the system:

$$T_{\mathrm{cw}} = \frac{1}{2}l^2\dot{\theta}^2(M+Mu+m) + \frac{1}{2}ml^2(\dot{\theta}-\dot{\phi})^2 - ml^2\dot{\theta}(\dot{\theta}-\dot{\phi})\cos\phi \tag{4.5}$$

Lagrangian:

$$L = T_{\mathrm{cw}} - V_{\mathrm{cw}}$$

For the stance leg angle θ, the Euler-Lagrange equation:

$$\frac{\mathrm{d}}{\mathrm{d}t}(\frac{\partial L}{\partial \dot{\theta}}) - \frac{\partial L}{\partial \theta} = 0$$

$$\begin{aligned}
\frac{\mathrm{d}}{\mathrm{d}t}(\frac{\partial L}{\partial \dot{\theta}}) &= l^2\ddot{\theta}(M+Mu+m) + ml^2(\ddot{\theta}-\ddot{\phi}) + ml^2\dot{\phi}(\dot{\theta}-\dot{\phi})\sin\phi - ml^2(\ddot{\theta}-\ddot{\phi})\cos\phi \\
&- ml^2(\ddot{\theta}\cos\phi - \dot{\theta}\dot{\phi}\sin\phi) \\
\frac{\partial L}{\partial \theta} &= (m-Mu+m)gl\sin\theta - mgl\sin(\theta-\phi)
\end{aligned}$$

Substituting $\frac{m}{M} = 0$ in the Euler-Lagrange equation and scaling time by $\sqrt{\frac{l}{g}}$, we get:

$$\begin{aligned}
(1+u)l^2\ddot{\theta} &= (1-u)gl\sin\theta \\
\ddot{\theta} &= \frac{1-u}{1+u}\sin\theta
\end{aligned} \tag{4.6}$$

For the swing leg angle ϕ, the Euler-Lagrange equation:

$$\frac{\mathrm{d}}{\mathrm{d}t}(\frac{\partial L}{\partial \dot{\phi}}) - \frac{\partial L}{\partial \phi} = 0$$

$$\begin{aligned}
\frac{\mathrm{d}}{\mathrm{d}t}(\frac{\partial L}{\partial \dot{\phi}}) &= -ml^2(\ddot{\theta}-\ddot{\phi}) - ml^2\dot{\theta}\dot{\phi}\sin\phi + ml^2\sin(\phi-\theta) \\
\frac{\partial L}{\partial \phi} &= ml^2(\dot{\theta}-\dot{\phi})\dot{\theta}\sin\phi + mglsin(\theta-\phi)
\end{aligned}$$

Dividing the Euler-Lagrange equation by m and scaling time by $\sqrt{\frac{l}{g}}$, we get:

$$\ddot{\theta} - \ddot{\phi} + \dot{\theta}^2\sin\phi - \ddot{\theta}\cos\phi + \sin(\theta-\phi) = 0$$

Using equation A.3 and an trigonometric identity, we obtain:

$$\ddot{\phi} = \ddot{\theta} + \dot{\theta}^2\sin\phi - \cos\theta\sin\phi + \frac{2u}{1+u}\sin\theta\cos\phi \tag{4.7}$$

Due to the hip torque used for actuation (equation 4.1), this equation is modified to:

$$\ddot{\phi} = \ddot{\theta} + \dot{\theta}^2\sin\phi - \cos\theta\sin\phi + \frac{2u}{1+u}\sin\theta\cos\phi - k_{\mathrm{f}}\phi \tag{4.8}$$

Tuned spring system

The tuned spring system is based on the concept of using a spring (elastic element) to provide an unloading force. According to the hypothesis presented in [43], the stiffness of the spring used for providing the unloading force can be tuned to minimize the inertia caused by the unloaded mass. The stiffness of this spring is given by:

$$k_{\mathrm{s}} = u\omega^2 M \quad \& \quad \omega = 2\pi c$$

where c is the cadence of the walking model at 0% BWU.
To provide unloading force when $y_1 = l$ (similar to a human standing still), the initial deflection of the spring should be:

$$\Delta l_0 = \frac{uMg}{k_s} = \frac{g}{\omega^2}$$

Thus, the deflection of the spring at a given time instant is given by:

$$\Delta y = \Delta y_0 + l - y_1 = \frac{g}{\omega^2} + l - l\cos\theta$$

Potential energy 'V_{ts}' of the system:

$$\begin{aligned} V_{ts} &= Mgy_1 + mgy_2 + \frac{1}{2}k_s\Delta y^2 \\ V_{ts} &= (M+m)gl\cos\theta - mgl\cos(\theta-\phi) + \frac{1}{2}\omega^2 uM(\frac{g}{\omega^2} + l - l\cos\theta)^2 \end{aligned} \tag{4.9}$$

Kinetic energy 'T_{ts}' of the system:

$$L = \frac{1}{2}(M+m)l^2\dot\theta^2 + \frac{1}{2}ml^2(\dot\theta-\dot\phi)^2 - ml^2\dot\theta(\dot\theta-\dot\phi)\cos\phi \tag{4.10}$$

Lagrangian:

$$L = T_{ts} - V_{ts}$$

For the stance leg angle θ, the Euler-Lagrange equation:

$$\frac{\mathrm{d}}{\mathrm{d}t}(\frac{\partial L}{\partial\dot\theta}) - \frac{\partial L}{\partial\theta} = 0$$

$$\begin{aligned} \frac{\mathrm{d}}{\mathrm{d}t}(\frac{\partial L}{\partial\dot\theta}) &= (M+m)l^2\ddot\theta + ml^2(\ddot\theta-\ddot\phi) - ml^2(\ddot\theta-\ddot\phi)\cos\phi + ml^2(\dot\theta-\dot\phi)\dot\phi\sin\phi - ml^2\ddot\theta\cos\phi + ml^2\dot\phi\dot\theta\sin\phi \\ \frac{\partial L}{\partial\theta} &= (M+m)gl\sin\theta - mgl\sin(\theta-\phi) + \omega^2 uMl^2\cos\theta\sin\theta - \omega^2 uMl^2\sin\theta - uMgl\sin\theta \end{aligned}$$

Substituting $m/M = 0$ in the Euler-Lagrange equation and scaling time by $\sqrt{\frac{l}{g}}$, we get:

$$\ddot\theta = (1-u)\sin\theta + \frac{l}{g}\omega^2 u(1-\cos\theta)\sin\theta \tag{4.11}$$

For the swing leg angle ϕ, the Euler-Lagrange equation:

$$\frac{\mathrm{d}}{\mathrm{d}t}(\frac{\partial L}{\partial\dot\phi}) - \frac{\partial L}{\partial\phi} = 0$$

$$\begin{aligned} \frac{\mathrm{d}}{\mathrm{d}t}(\frac{\partial L}{\partial\dot\phi}) &= -ml^2(\ddot\theta-\ddot\phi) + ml^2\ddot\theta\cos\phi - ml^2\dot\phi\dot\theta\sin\phi \\ \frac{\partial L}{\partial\phi} &= ml^2(\dot\theta-\dot\phi)\dot\theta\sin\phi + mgl\sin(\theta-\phi) \end{aligned}$$

Dividing the Euler-Lagrange equation by ml^2 we get:

$$\ddot\phi - \ddot\theta + \ddot\theta\cos\phi - \dot\phi\dot\theta\sin\phi = \dot\theta(\dot\theta-\dot\phi)\sin\phi + \frac{g}{l}\sin(\theta-\phi)$$

Scaling time by $\sqrt{\frac{l}{g}}$ and using equation A.3 and an trigonometric identity, we obtain:

$$\ddot\phi = \ddot\theta + \dot\theta^2\sin\phi - \cos\theta\sin\phi - u(1 + \frac{l}{g}\omega^2(1-\cos\theta))\sin\theta\cos\phi \tag{4.12}$$

Due to the hip torque used for actuation (equation 4.1), this equation is modified to:

$$\ddot\phi = \ddot\theta + \dot\theta^2\sin\phi - \cos\theta\sin\phi - u(1 + \frac{l}{g}\omega^2(1-\cos\theta))\sin\theta\cos\phi - k_f\phi \tag{4.13}$$

Optimization

The SW model was optimized at each BWU condition in order to achieve a stable gait cycle. The following parameters were optimized: 1) step length as it governs the initial value of θ, 2) ω which determines the initial value of ϕ through the relation $\phi = \omega\theta$ and 3) k_f which is the torsional stiffness of the hip spring. These parameters were optimized using the Matlab function `fmincon` to minimize the cost function $C = \frac{1}{d}$, where d is the distance traveled by the model during one iteration. By minimizing C, the distance traveled was maximized. The starting values for the optimization procedure were the parameter values that produced a stable gait at 0% BWU. However, the results shown by using this cost function were inconsistent, possibly due to the existence of local minima and hence the cost function was modified to consider the state of the model. This cost function was:

$$C = q_{\text{err}} + s_{\text{err}} \quad \text{where} \quad q_{\text{err}} = \frac{\sum_{i=1}^{n-1} ||[q_{i+1} - q_i]||_1}{n-1} \quad \& \quad s_{\text{err}} = \frac{\sum_{i=1}^{n-1} [s_{i+1} - s_i]}{n-1}$$

where $q_i = [\theta \quad \dot{\theta} \quad \phi \quad \dot{\phi}]$ is the state vector at the beginning of step i, s_i is the step length for step i and n is total number of steps simulated. This cost function was aimed at minimizing the error between the state vectors in consecutive steps, thus making it more likely for the model to achieve a stable gait. This cost function led to consistent results and was implemented for the final simulations.

4.2. Spring-loaded inverted pendulum model

This section presents a basic description of the bipedal spring-loaded inverted pendulum (SLIP) model by [44], followed by the derivation of the equations of motion for the three BWS systems.

Model description

The SLIP model (Figure 4.2) represents the legs as two linear, massless springs of equal stiffness k and length l_0. The springs influence the dynamics of the model only during stance phase when the spring force counteracts the gravitational force. Each spring acts independently and it has no physical meaning during the swing phase. The spring describes a kinematic touchdown condition during swing, $y_{\text{TD}} = l_0 \sin\alpha_0$ given by the rest length of the spring l_0 and the fixed leg orientation α_0 (constant angle-of-attack) with respect to gravity. Heelstrike indicates the transition of swing to stance and the stance to swing transition occurs when the springs reaches the rest length during lengthening. The SLIP model combines the basic dynamics of running and walking gaits into a single mechanical system and can be used as a template for more complex models of human gait. Please refer to [44] for more details.

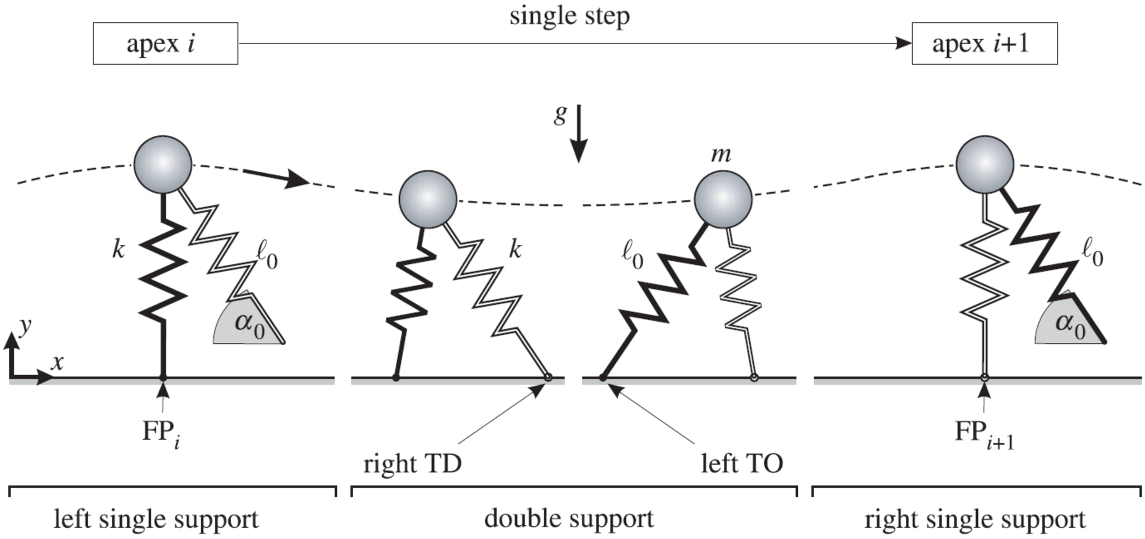


Figure 4.2: Schematic of the SLIP model. The schematic shows a single step which begins at the highest COM at apex i, includes the double support and ends at the next apex $i+1$ in right single support. The position of the foot point in single support is denoted by FP. Figure reprinted from [44]

Model parameters for the SLIP model are:

- 'M' is the mass at hip
- (x, y) is the location of the hip mass
- Massless spring legs with stiffness k and rest length l_0
- Stance foot is at the origin
- Constant angle of attack α_0
- $\phi(t)$ is the angle between the swing and the stance legs
- β is the percentage of BWU provided and u is equal to $U/100$

Equations of motion for the centre of mass (x, y) are –

1. Initial left leg support:

$$\begin{aligned} m\ddot{x} &= Px \\ m\ddot{y} &= Py - mg \end{aligned} \tag{4.14}$$

2. Intermittent double leg support:

$$\begin{aligned} m\ddot{x} &= Px - Q(d-x) \\ m\ddot{y} &= Py + Qy - mg \end{aligned} \tag{4.15}$$

3. Final right leg support:

$$\begin{aligned} m\ddot{x} &= -Q(d-x) \\ m\ddot{y} &= Qy - mg \end{aligned} \tag{4.16}$$

where FP is foot point of a stance spring and

$$\begin{aligned} P &= k(\frac{l_0}{\sqrt{x^2+y^2}} - 1) \\ Q &= k(\frac{l_0}{\sqrt{(d-x)^2+y^2}} - 1) \\ d &= \mathrm{FP}_{i+1,x} - \mathrm{FP}_{i,x} \end{aligned}$$

Constant force

A term representing the constant vertical unloading force ($F_u = uMg$) is added to the original equations [44]. The equations for $\ddot{x}$ remain the same as above since gravity only affects the vertical motion. The resulting equations of motion for $\ddot{y}$ are:

$$\text{Initial left leg support:} \quad m\ddot{y} = Py - m(1-u)g \tag{4.17}$$

$$\text{Intermittent double leg support:} \quad m\ddot{y} = Py + Qy - m(1-u)g \tag{4.18}$$

$$\text{Final right leg support:} \quad m\ddot{y} = Qy - m(1-u)g \tag{4.19}$$

Counterweight system

Mass of the counterweight is 'uM' where where 'u' represents the amount of BWU provided as a ratio of body mass and M is the mass at the hip. The equations for $\ddot{x}$ remain the same as above since BWS is assumed to affect only the vertical motion. The resulting equations of motion for $\ddot{y}$ are:

$$\begin{aligned} \text{Initial left leg support:} \quad m\ddot{y} &= Py - m\frac{(1-u)}{1+u}g & (4.20) \\ \text{Intermittent double leg support:} \quad m\ddot{y} &= Py + Qy - m\frac{(1-u)}{1+u}g & (4.21) \\ \text{Final right leg support:} \quad m\ddot{y} &= Qy - m\frac{(1-u)}{1+u}g & (4.22) \end{aligned}$$

Tuned spring system

The tuned spring system is based on the concept of using a spring (elastic element) to provide an unloading force. According to the hypothesis presented in [43], the stiffness of the spring used for providing the unloading force can be tuned to minimize the reflected inertia caused by the unloading system. The stiffness of this spring is given by:

$$k_s = u\omega^2 M \quad \& \quad \omega = 2\pi c$$

where c is the cadence of the walking model at 0% BWU.
To provide unloading force at the start of the gait ($y = y_0$), the initial deflection of the spring should be:

$$\Delta l_0 = \frac{uMg}{k_s} = \frac{g}{\omega^2}$$

Thus, the deflection of the spring at a given time instant is given by:

$$\Delta y = \Delta l_0 + y_0 - y$$

The equations for $\ddot{x}$ remain the same as above since BWS is assumed to affect only the vertical motion. The resulting equations of motion for $\ddot{y}$ are:

$$\begin{aligned} \text{Initial left leg support:} \quad m\ddot{y} &= Py - mg + k_s\Delta y & (4.23) \\ \text{Intermittent double leg support:} \quad m\ddot{y} &= Py + Qy - mg + k_s\Delta y & (4.24) \\ \text{Final right leg support:} \quad m\ddot{y} &= Qy - mg + k_s\Delta y & (4.25) \end{aligned}$$

Optimization

The SLIP model was optimized for each BWU condition in order to achieve a stable gait cycle. The following parameters were optimized: 1) initial horizontal velocity of the COM $\dot{x}$, 2) the touchdown angle (angle-of-attack) of the swing leg α_0 and 3) k which is the stiffness of the leg spring. These parameters were optimized using the Matlab function `fmincon` to minimize the cost function $C = \frac{1}{d}$, where d is the distance traveled by the model during one iteration. By minimizing C, the distance traveled was maximized. The starting values for the optimization procedure for each BWU condition were the parameter values that produced a stable gait at 0% BWU.

4.3. Muscle-reflex model

This section presents a basic introduction to the Muscle-reflex gait model [45], followed by the Simscape implementation of the BWS strategies.

Model description

The Muscle-reflex (MR) model is conceptually based on the bipedal spring-mass model [44] and encodes the principles of legged locomotion using autonomous muscle reflexes. These muscle reflexes have been tuned manually with the goal of achieving a human-like gait for the model. It represents the human body using two three-segmented legs and a 'trunk', which represents head, arms and trunk (Figure 4.3). Seven Hill-type muscles [46] are used to actuate each leg. The soleus and lumped vasti group (VAS) muscles are driven with positive force feedbacks F+ to produce the compliant behavior presented in [44] (Figure 4.3B). The biarticular gastrocnemius (GAS) muscle is driven using positive force feedback F+ and the VAS is inhibited when the knee extension angle is greater than 170°. This prevents the overextension of the knee (Figure 4.3C). Positive length feedback L+ of the tibialis anterior (TA) muscle prevents the overextension of the ankle (Figure 4.3C). However, a negative force feedback F- from the soleus muscle is used to suppress this positive feedback during normal stance conditions.

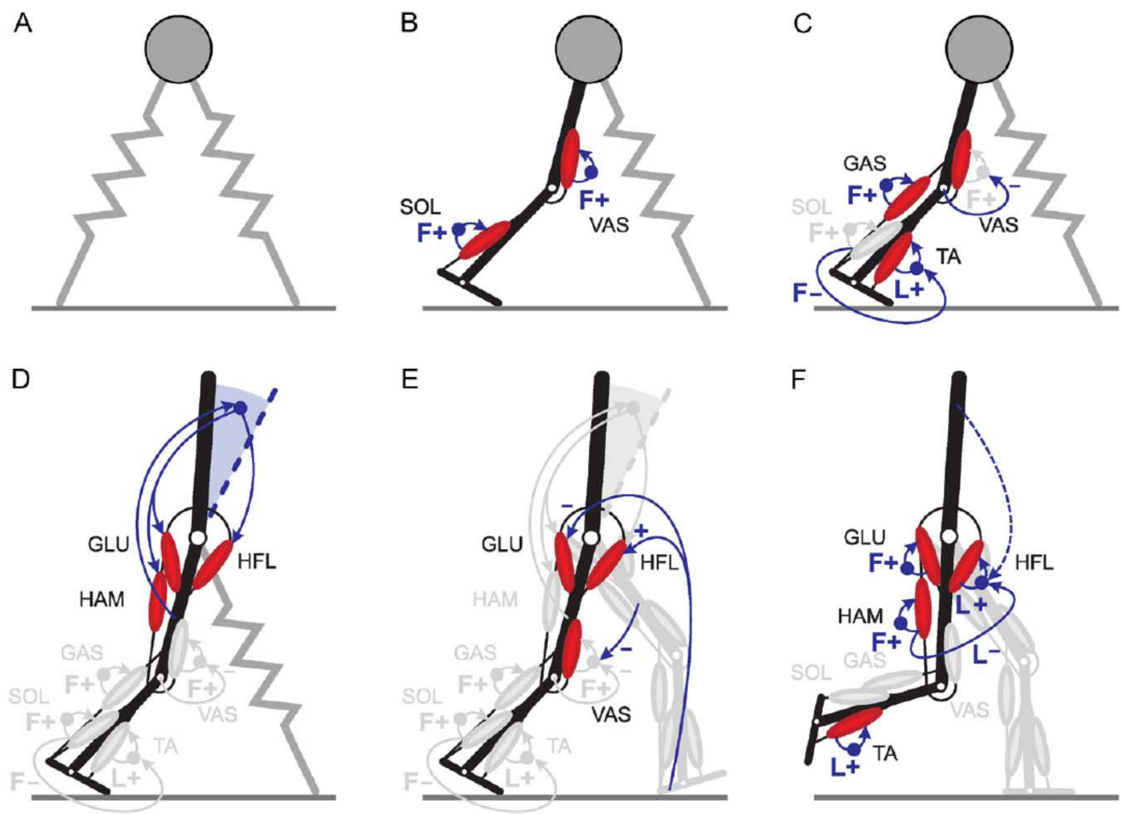


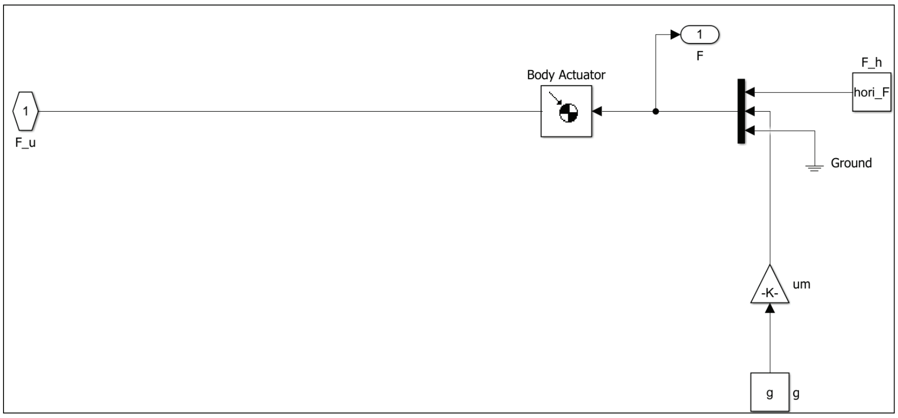
Figure 4.3: Schematic of the MR model. Figure reprinted from [45]

The hip flexors muscle group (HFL) and the co-activated hip extensor muscles, gluteal (GLU) muscle group and hamstring muscle group), propel the trunk into a reference lean position (Figure 4.3D). The reflexes for the trunk are modulated according to the load borne by the stance leg. The biarticular HAM is also utilized to stop the overextension of knee which can result from the hip extensor torques. Swing is initiated (Figure 4.3E) via adding or subtracting a constant stimulation to HFL or GLU respectively and through the suppression of VAS in proportion to the load borne by the leading (other) leg. The positive length feedback L+ of HFL enables the leg swing and the L- of HAM is used to finally suppress the swing. At the end of swing, the leg is retracted and then straightened by the F+ positive force feedback of HAM and GLU. Before landing, the L+ of TA (Figure 4.3F) moves the ankle joint into a flexed configuration.

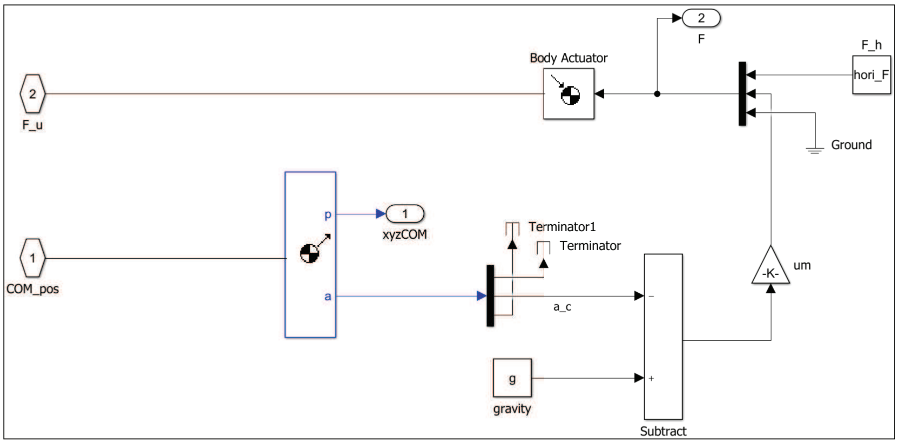
The MR model does not contain a central pattern generator to activate the muscle-reflexes. It switches between the stance and swing reflexes for the right and left legs using the ground contact response of the sensors at the heel and ball of each foot. These sensors emulate the mechanoreceptors in the foot, which are considered to be important for the modulation of phase transitions [45]. Similarly, the MR model also encodes and emulates other biological principles of legged locomotion like compliant behaviour of legs, swing-leg retraction, ballistic motions of lower legs in swing phase, indirect generation of ankle push-off motion, prevention of joint overextension in segmented chains, etc [45]. Manual tuning of the muscle-reflexes not only leads to the emergence of human walking dynamics and kinematics, the MR model adapts to slopes and tolerates ground disturbances. For further details, please refer to [45].

Implementation of BWS strategies

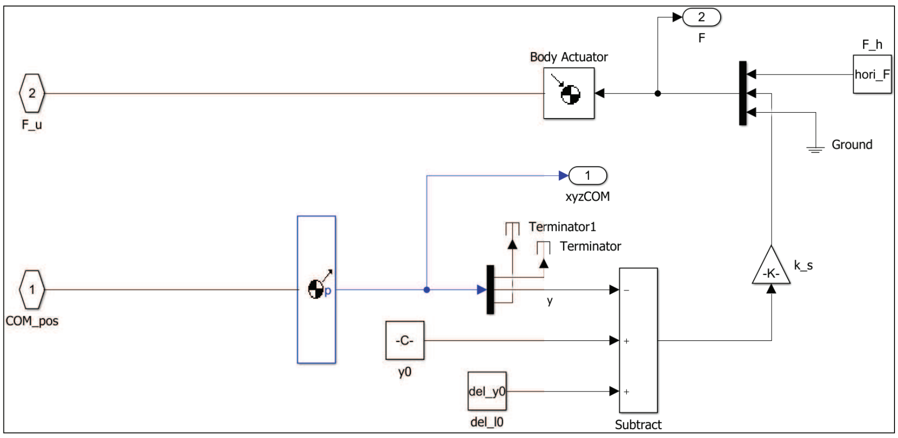
Equations (20-22) from the main paper describe the unloading force for the Constant force (CF), Counterweight (CW) and Tuned spring (TS) BWS strategies respectively. This unloading force was applied at the centre of mass of the upper body. A separate Simscape sub-system was created to emulate each BWS strategy since the original model [45] was implemented in Simscape. These sub-systems are shown in Figure 4.4.



(a) Schematic for the Constant force (CF) BWS strategy



(b) Schematic of the Counterweight (CW) BWS strategy



(c) Schematic of the Tuned spring (TS) BWS strategy

Figure 4.4: F_u is the unloading force, u is the proportion of unloading, m is the total mass of the body, a_c (Subfigure b) is the downward acceleration of the attachment point of the BWS system (COM of the upper body), y0 (Subfigure c) is the vertical position of the COM of the upper body at time $t = 0$, y (Subfigure c) is the vertical position of the COM of the upper body at time t, del_l0 (Subfigure c) is the elongation of the BWS spring at time $t = 0$ and g is the gravitational acceleration. The horizontal force F_h is set to 0 since we are only concerned with the vertical unloading force

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