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How gravity and muscle action control mediolateral center of mass excursion during slow walking: A simulation study

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ABSTRACT

Maintaining mediolateral (ML) balance is very important to prevent falling during walking, especially at very slow speeds. The effect of walking speed on support and propulsion of the center of mass (COM) has been focus of previous studies. However, the influence of speed on ML COM control and the associated coupling with sagittal plane control remains unclear. Simulations of walking at very slow and normal speeds were generated for twelve healthy subjects. Our results show that gluteus medius (GMED) contributions to ML stability decrease, while its contributions to sagittal plane accelerations increase during very slow compared to normal walking. Simultaneously the destabilizing influence of gravity increases in ML direction at a very slow walking speed. This emphasizes the need for a tight balance between gravity and gluteus medius action to ensure ML stability. When walking speed increases, GMED has a unique role in controlling ML acceleration and therefore stabilizing ML COM excursion. Contributions of other muscles decrease in all directions during very slow speed. Increased contributions of these muscles are therefore required to provide for both stability and propulsion when walking speed increases.

1. Introduction

A high incidence of falls is an important problem in modern society. In addition to the high associated cost on health care, these falls also have a major impact on the quality of life [1]. Many of these falls occur during walking.

To maintain balance during walking, control of the mediolateral (ML) motion of the center of mass (COM) is crucial. Accordingly, deviations of the gait pattern in the ML direction are often suggested to be a valid predictor of falls. Previous studies suggested step width (SW) and especially SW variability to be related to fall risk [2]. During steady-state gait, older adults reduce the ML COM accelerations as a compensatory mechanism to improve ML stability [3].

These modifications in COM movement and therefore its control are likely to become important at very slow speeds. Den Otter et al. [4] found specific bursts of muscle activity at very slow speeds and argued that these might be attributed to increased demands on postural stability. The direct relation to ML COM control still needs to be further investigated. As several studies found indeed that slower walking in elderly is associated with an increased risk of falls [5], it is highly relevant to understand how slow walking affects the ML COM control. To date, the relation between the ML and sagittal plane COM control remains unclear. In the sagittal plane, support and progression are coupled and similar muscles contribute to both in parallel. Principle component analysis of experimental electromyography (EMG) patterns shows that a reduced set of muscle activation modules can generate both support and propulsion [6]. This is also confirmed using muscle driven simulations [7–9]. Furthermore muscle contributions to both progression and support generally decrease proportionally when walking speed decreases [10,11].

Extending these insights to the ML COM control is less clear. Some experimental studies on posture suggest that ML and anterior–posterior (AP) motions have a coupled control [12]. In

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contrast, other researchers believe that COM control in the sagittal plane and ML stabilization are independent of each other during quiet stance [13] and balance perturbations [14]. Proof also exists for an independent control during gait, with decreasing walking speed, local dynamic stability in AP and vertical direction is enhanced, but stability in ML direction reduced [15]. This indicates that separate control of stability is required in different directions. Likewise, analysis of passive walking models suggests that during walking ML stability is actively controlled, while passive stability is provided in the sagittal plane [16]. Simulation studies of steady-state walking [17] investigated how individual muscles control ML balance. They concluded that muscles responsible for sagittal plane accelerations also accelerated the COM laterally (i.e. away from the midline) and that COM stability in all directions is controlled by similar muscle groups. However, a recent study [18] showed that control of ML accelerations of the COM requires additional synergies to the ones controlling the COM in the sagittal plane. Hence, these studies do not allow to uniformly conclude on sagittal plane versus ML COM control. Previous simulation studies [10,11] mainly explored the effect of speed on the muscle contributions to COM control in the sagittal plane. The study of Pandy et al. [17] investigated ML COM stability, but only reported one single speed. Only recently, a study of John et al. [19] explored the effect of walking speed on muscle contributions to the control of ML body motion. However, slowest speeds reported in their study are still largely above the walking speeds previously reported in for instance stroke patients and the subjects involved were children. Therefore, when exploring the role of individual muscle contributions to ML and sagittal plane COM control it seems advantageous to investigate the trade-off between sagittal plane and ML control also at very slow walking speeds and in an adult population.

In the present study, the effect of walking speed on ML and sagittal plane COM control is investigated using simulations of walking at normal and very slow speed. More particularly, we investigate if a decrease in speed affects similarly the muscle contributions in ML, vertical and AP direction. If a common control exists for the different planes, one would expect that with decreasing speed the observed decreases in muscle action (EMG) would have similar effects on the muscle contributions to COM accelerations in the various planes. However, if changes in muscle contributions to COM acceleration are specific for a given plane this would argue for separate control. One important muscle to consider is the gluteus medius (GMED). This muscle decreases EMG activity with decreasing gait speed [20]. Based on a simulation study, Liu et al. [10] concluded that GMED contributions to support are 'relatively constant across walking speeds'. However, a significant difference is reported between free and slow walking speed, with higher GMED contributions for the slow speeds.

Based on the assumption that the functional demands for progression decrease whereas stability remains equal or also decreases when walking at very slow speeds, it is hypothesized that muscles that act mainly in the sagittal plane will decrease their contributions in this plane, but maintain their ML contribution. In contrast, muscles preferentially contributing to ML stability such as GMED mainly decrease their contribution to ML accelerations of COM but maintain their contributions in the sagittal plane. This will therefore confirm an uncoupling of the COM control in the ML and sagittal plane.

2. Methods

2.1. Experimental data

We collected three-dimensional kinematic and kinetic data of twelve healthy subjects (age: 25.8 ± 4.0 years, weight: 71.1 ± 8.9 kg, leg length: 0.9 ± 0.04 m) walking on treadmill at a speed of 1 km/h (very slow) and 4 km/h (normal). The speed of 1 km/h was chosen because it is at the low end of the range of preferred speeds in stroke patients [21].

All subjects gave their informed consent prior to data collection and the experimental protocol was approved by the local ethical committee. Marker (active infrared LEDs) trajectories were collected at 100 Hz using a two-beam camera system (Krypton, Nikon Metrology NV, Belgium). The marker protocol consisted of six technical clusters and 16 additional individual markers; this protocol was described more detailed in a previous paper [22]. Ground reaction forces (GRF) and torques were measured at 1000 Hz using a force-plate instrumented split-belt treadmill (Forcelink, The Netherlands). EMG data were collected bilaterally at 1000 Hz for tibialis anterior, gastrocnemius lateralis, soleus (SOL), vastus lateralis, rectus femoris (RF), biceps femoris and semitendinosus using a wireless EMG system (Zero-wire EMG, Aurion, Italy). As part of the post-processing, the raw EMG signal was band-pass filtered between 10 and 500 Hz using a fourth order digital Butterworth filter and RMS was calculated using a 50 ms time window.

2.2. Simulations

Subject-specific simulations of both walking speeds were generated using a dedicated workflow in OpenSim [23]. In a first step, data of a static trial was used to scale a generic musculoskeletal model [24] to match the anthropometry of the subject. This generic model consists of 27 degrees of freedom (Supplemental Table 1). The leg and trunk joints were actuated by 92 Hill-type muscle-tendon units and the arms were driven by torque actuators. An in house developed Kalman smoothing algorithm [25] calculated the joint angles that minimize the difference between experimental and model markers. A residual reduction algorithm [26] reduced dynamic inconsistencies between the model kinematics and the measured GRF (Supplemental Fig. 1). Computed Muscle Control computed the most optimal muscle excitations patterns required to track the experimental walking task [26]. The calculated muscle activations were compared to the subject's measured EMG and visually verified (Supplemental Fig. 2). Simulated kinematics and kinetics closely tracked experimental kinematics and kinetics (Supplemental Figs. 3 and 4).

2.3. Analysis muscle function

A perturbation analysis [8] computed the AP, vertical and ML COM accelerations induced by a specific muscle or gravity. The force of each muscle or gravity was subsequently increased with 1 N and the equations of motion were integrated forward over a time window of 0.03 s to evaluate the effect on the COM. To allow changes in the GRF and moments during the perturbation, linear and torsional springs were added between the model's feet and the floor. We verified the validity of the simulations by comparing the summed muscle contributions to the COM acceleration with the COM accelerations in the reference simulation (Supplemental Fig. 5).

Contributions of muscles and gravity were calculated over the entire gait cycle, which was further divided into subphases (Figs. 2 and 3). While most simulation studies [8,10,11,17] consider head, arms and trunk as a single rigid body, this study used a model that included separate arms segments. This allowed us to identify the contribution of the arm dynamics to the COM accelerations.

In a post processing step, positive and negative contributions were separately averaged over each subphase and subsequently over all subjects. To simplify data analysis, contributions of smaller muscles with similar function were summed.

We performed a Wilcoxon's test for matched pairs (Statistica, Statsoft Inc., USA) to test for significant differences in average contributions between normal and very slow walking.

3. Results

3.1. Total muscle contributions versus gravity

At a very slow speed, COM acceleration decreased (Fig. 1B) mainly due to a decrease in the total muscle contribution (Fig. 2, SUM). Gravity on the other hand showed mixed results: While the downward acceleration due to gravity decreased at a very slow speed, the contribution to anterior acceleration increased (Fig. 2, GRAVITY) during the entire gait cycle. Similarly, contributions to lateral acceleration during single stance and to medial acceleration during swing increased at a very slow speed.

3.2. Individual muscle contributions

At both speeds, plantarflexors (GAS and SOL), quadriceps (RF and VAS), DF and GMED contributed most to the AP and vertical COM acceleration during stance. Hip abductors (GMED and GMIN), ILIPSO, plantarflexors and VAS were the main contributors to ML accelerations (Fig. 3). The influence of the arm movement is negligible during both speeds (maximum arm contributions <0.1 m/s^2).

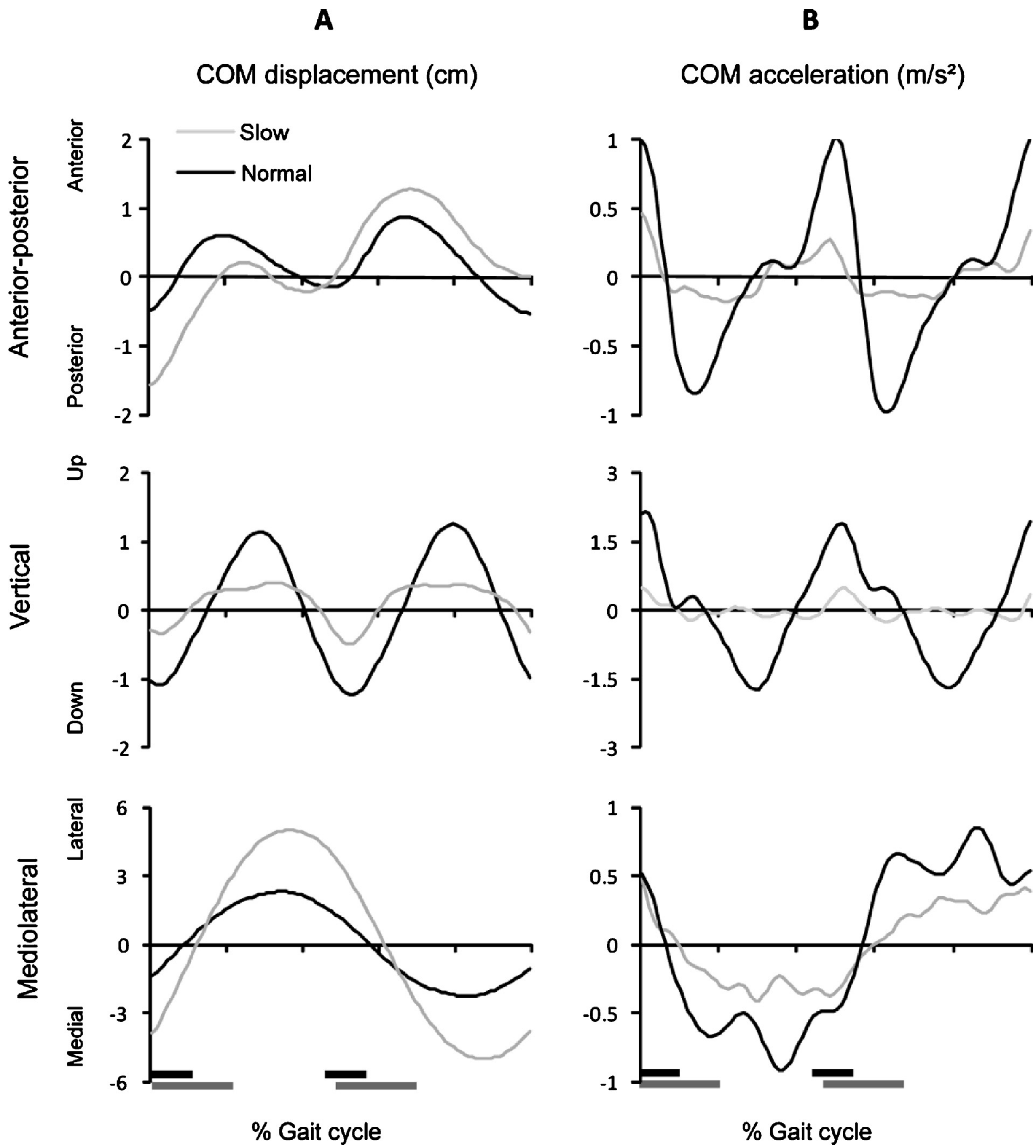

Fig. 1. Averaged ($n = 12$) displacements (A) and accelerations (B) of the center of mass (COM) over the gait cycle during slow and normal walking speed. Mediolateral center of mass excursions increased substantially in slow compared to normal walking speeds; center of mass accelerations over the gait cycle decreased significantly for all directions with decreasing walking speed. Shaded bars indicate periods of double-limb support during slow (gray) and normal (black) walking speed.

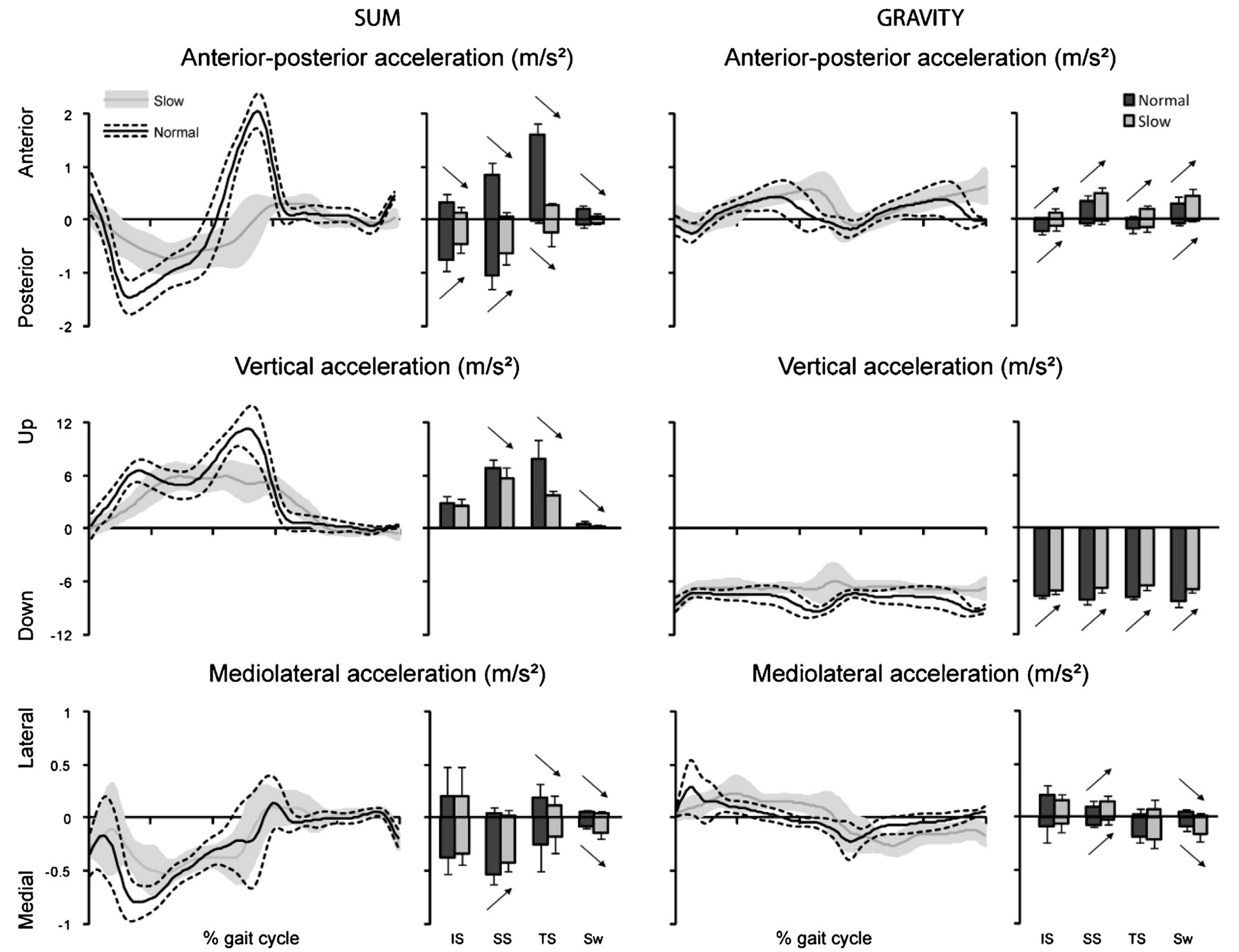

Fig. 2. Contributions of summed right leg muscles (SUM, left pane) and gravity (right pane) to the anterior–posterior, vertical and mediolateral accelerations of the center of mass during slow and normal walking speeds. (A) Absolute contributions over the gait cycle. (B) Contributions averaged over the sub phases of the gait cycle (n = 12, mean + 1 SD). A positive sign indicates contributions to respectively anterior, upward or lateral acceleration of the COM, a negative sign indicates contributions to respectively posterior, downward or medial acceleration of the COM. If contributions of both signs are present, it reflects a reversed contribution within that particular phase of the gait cycle. Initial stance was from ipsilateral heel contact to contralateral toe off. Single stance was from contralateral toe off to contralateral heel contact, terminal stance was from contralateral heel contact to ipsilateral toe off and swing form ipsilateral toe off to ipsilateral heel contact. Statistically significant differences ($p < 0.05$) are indicated with ↑ or ↓.

At initial stance, contributions from HAM, VAS and GMAX to anterior–posterior, upward and lateral (i.e. away from the midline) acceleration of the COM decreased significantly at a very slow speed compared to normal speed. However, contribution from GMED to support (i.e. upward acceleration) increased at a very slow speed. In the other muscles (GAS, SOL and DF), no significant differences were found between the different walking speeds.

During single stance, contributions of GAS to propulsion (i.e. anterior acceleration), support and lateral acceleration decreased during very slow compared to normal speed. Similarly, GMAX contributions to support and medial acceleration (i.e. toward the midline) decreased. On the other hand, increased contributions were found from GMED to propulsion and support, while contributions to medial acceleration decreased. ILIPSO and QUAD contributions remained relatively constant between the two speeds. Increased contributions were found for DF to propulsion and SOL to posterior acceleration.

In terminal stance, contributions of plantarflexors (GAS and SOL) to propulsion, support and lateral acceleration all decreased. ILIPSO contribution to propulsion, downward and lateral acceleration also decreased, similar as GMIN contributions in the opposite directions. Similar as in previous phases, contributions of hip abductor GMED increased in the sagittal plane, but decreased in medial direction.

During swing, contributions of most muscles were small at both speeds. The contribution of ILIPSO to lateral acceleration decreased during very slow walking. In contrast, an increase was found of DF contribution to propulsion at very slow speed.

In general, at very slow speed, muscles contributions to mediolateral acceleration either decreased or remained constant compared to normal speed. A similar pattern was found for muscles contributions to sagittal plane COM acceleration. GMED consistently increased its contribution to sagittal plane COM control.

4. Discussion

In the literature, correlations were found between an increase of fall risk, ML deviations of the COM and slower walking speeds [2,4]. The purpose of this study was therefore to gain more insight in the influence of speed on ML COM control, and additionally how this ML control is coupled with sagittal plane control. Therefore,

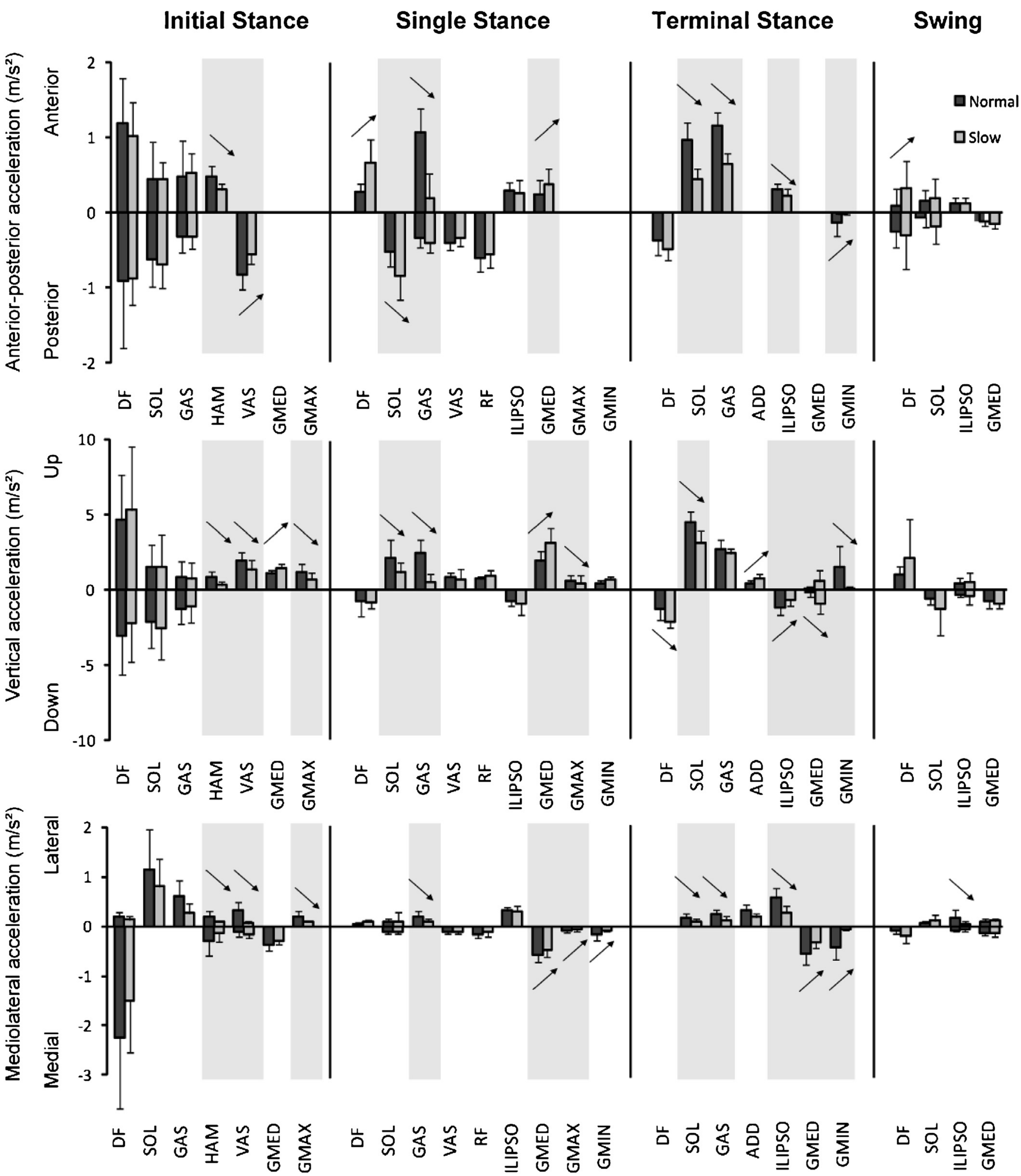

Fig. 3. Contributions of individual muscles to anterior–posterior (top), vertical (middle) and mediolateral (bottom) accelerations of the center of mass, averaged over the initial stance, single stance, terminal stance and swing ($n = 12$; mean + 1 SD). GMAX consists of the three parts of gluteus maximus, GMED consists of the three parts of the gluteus medius. GMIN includes the different parts of gluteus minimus. ILIPSO consists of iliacus and psoas and ADD consists of adductor longus, brevis and magnus. HAMS combines semimembranosus, semitendinosus, and biceps femoris long head. VAS consists of vastus lateralis, intermedius and medialis. GAS consists of the medial and lateral parts of the gastrocnemius; dorsiflexors (DF) consist of tibialis anterior, extensor digitorum and extensor hallucis. Statistically significant differences ($p < 0.05$) are indicated with ↑ or ↓. As contributions from ipsilateral and contralateral side are symmetric, only ipsilateral contributions are shown.

we generated subject specific 3D simulations of gait at very slow and normal speeds.

4.1. Mediolateral control of the COM

When walking speed decreases from normal to very slow, resulting medial COM accelerations decreased during single stance. Two factors contribute to this decrease:

Firstly, during very slow walking, *gravity's* destabilizing contribution to *lateral* acceleration (away from the midline) increases. The increased contribution of gravity to lateral COM acceleration (Fig. 2) is indeed associated with increased ML COM displacement (Fig. 1A). The higher destabilizing action of gravity might well be the reason why even healthy individuals increase their step width [27] at very slow speeds, as this counteracts the influence of gravity (cf. Pandy et al. [17]).

Secondly, during very slow walking, both *net* and *individual muscle contributions* to medial COM acceleration decrease significantly compared to normal walking. Especially decreased abductor (i.e. GMIN and GMED) contributions (Fig. 3) to medial COM acceleration are responsible for this decrease.

The excessive lateral COM movement (Fig. 1A) during slow walking speeds, also reported in previous studies [27], can therefore be related to a combined decrease of the restraining muscle action in medial direction and increased destabilizing GRAV action in lateral direction.

4.2. Coupling of control in the different planes

4.2.1. Gravity

Similar to its role in controlling ML acceleration, an increase of gravity contribution to forward propulsion was found at slow compared to normal walking (Fig. 2). In combination with a large decrease in muscle propulsive action, this leads to a very high relative contribution of gravity to forward propulsion during slow walking. To compensate for this, a less pronounced decrease (in single stance) and even increase (in terminal stance) of total muscle contributions to posterior acceleration are found during slow walking. Hereby, a higher backward tilting moment is induced to maintain balance in the sagittal plane.

In contrast with AP and ML directions, a decreased contribution of gravity to downward acceleration at very slow speed is found in this study (Fig. 2), which is consistent with an increase of skeletal alignment [10]. An uncoupling of gravity's contributions to COM control in the different directions therefore exists.

4.2.2. Muscles

For most muscles a coupled control between sagittal plane and ML stability was shown, with muscle contributions that decreased at very slow compared to normal speed in both the sagittal plane (cf. Liu et al. [10]) and mediolateral direction (cf. John et al. [19]). However, the decrease of ML GMED contributions combined with the increase of sagittal plane contributions during very slow walking, suggests that an uncoupling of GMED muscle function underlies the change in walking speed. This confirms the assumption of a separate control mechanism for sagittal plane and ML balance control [15,16].

Further analysis of our results (Fig. 4) shows an uncoupling in acceleration potential of GMED, i.e. the acceleration per Newton of force. The acceleration potential of GMED in AP (Sw: +90%) and UD (IS: +31%, SS: +46%, TS: +384%, Sw: +103%) direction increased at very slow compared to normal walking speed, while the potential in ML (SS: −36%, TS: −37%) direction decreased. The force of GMED is the same in all directions. Therefore, the uncoupling of total of GMED contribution between the different planes, i.e. the product of the total muscle force with the potential acceleration, mainly results from the uncoupling of acceleration potential in the different directions.

The reduced ML potential of GMED at lower speeds may potentially threaten ML stability as less ML stability will be provided at slow walking speed. As contribution of GRAV to ML acceleration increases a critical balance needs to be satisfied to restrict excessive lateral motion of the COM during very slow walking.

Our results are in agreement with previous speed-dependent changes found by Liu et al. [10] and John et al. [19]. Similar as in Liu et al. [10], contributions of GMAX, HAM and VAS to support decreased at our very slow compared to the normal walking speed during early stance. During late stance, a decrease was found in soleus contribution to support. Likewise, GMED contribution increased instead of decreased at very slow speed. Our results on muscle contribution in antero-posterior direction also agree with Liu et al. [10]; VAS contribution to posterior acceleration decreased during early stance, and SOL contribution to propulsion decreased during late stance. In agreement with John et al. [19], we found decreased contributions of VAS and GMAX to lateral accelerations of the COM at initial stance during very slow walking. However, no significant changes were found in initial stance contributions to medial accelerations. GMED contributions to medial accelerations decreased in both single and terminal stance. A similar decrease of SOL contribution to lateral acceleration during TS was found. To the best of our knowledge, this study is the first to confirm decreased GAS contribution during TS to acceleration of COM in all three directions.

4.3. Methodological considerations/limitations

A potential limitation of our study is that subjects walked on a split-belt treadmill. No significant differences were found in SW

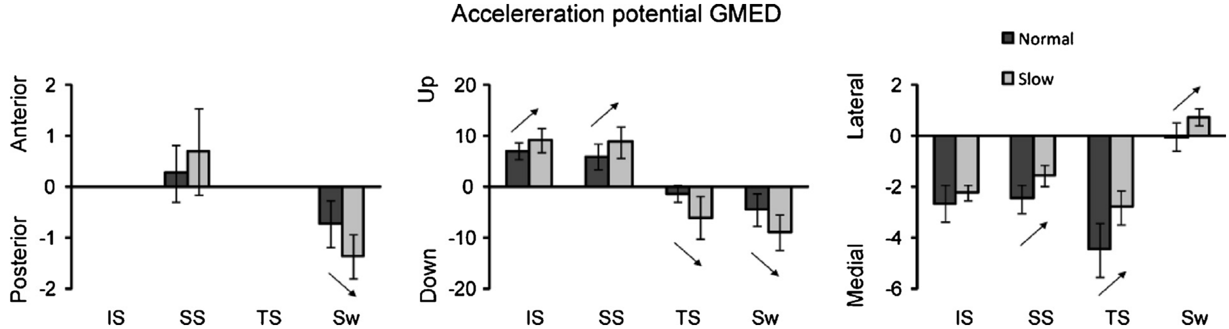

Fig. 4. Acceleration potential ($10^{-3} \times m/s^2 \times 1/N$) of gluteus medius (GMED) in the different directions, averaged over the initial stance (IS), single stance (SS), terminal stance (TS) and swing (Sw) (n = 12; mean ± 1 SD). GMED acceleration potentials are only reported in phases with significant total GMED contributions. Statistically significant differences ($p < 0.05$) are indicated with ↑ or ↓.

between slow and normal walking (0.22 ± 0.03 m vs. 0.23 ± 0.03 m, $p = 0.64$), while SW was expected to increase during slow compared to normal walking. This might be due to the use of a split-belt treadmill since the gap in between the belts might slightly bias the base of gait [28] compared to single-belt walking. However, the fact that there was no change in SW may actually be an advantage for the present study. Indeed the increase in SW related to walking speed found in overground walking, makes it difficult to evaluate the isolated effect of speed. The unchanged SW allows evaluating the separate effect of speed on the ML acceleration of the COM, without any confounding effects of SW. Furthermore, no significant differences were found in frontal plane joint moments, ML GRF or GMED EMG activity during the stance phase in split-belt treadmill walking compared to overground walking in the literature [29]. Frontal plane dynamic stability was also insensitive to overground or treadmill walking [30].

In this study, contributions of centrifugal, Coriolis and inertial forces were not taken into account. However, Anderson and Pandy [7] found that contributions to support during stance are relatively small. Furthermore, Pandy et al. [17] showed contributions from centrifugal and Coriolis forces to ML accelerations to be small compared to muscle contributions, especially during single stance.

4.4. Clinical implications

Slow walking seems to be advantageous for older people, because it requires less muscle force. However, while COM acceleration in ML direction decreases compared to normal speed, this is accompanied by an increase in ROM of the COM. This increased ROM induces higher demands on muscle coordination to keep equilibrium between gravity and muscle contributions. In elderly people, a reduction of their coordination capacity might jeopardize their ability to maintain balance at lower speeds. This will result in an increase in variability and might eventually lead to falls.

In conclusion, at very low walking speeds a vicious circle might arise in which increasing ROM and increasing destabilizing acceleration of gravity reinforce each other, eventually leading to falls. A tight coordination between the action of GMED toward the mid-line and lateral action of gravity away from the mid-line is therefore crucial to maintain stability. Increased contributions of most muscles are important for both stability and propulsion when walking speed increases. However, GMED has a unique role in controlling ML acceleration and therefore stabilizing ML COM excursion.

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Conflict of interest statement

No conflict of interest, financial or otherwise, is declared by the authors.

Appendix A. Supplementary data

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.gaitpost.2013.06.004.

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