Dynamic Balance During Human Movement: Measurement and Control Mechanisms

Richard R. Neptune[1]
Walker Department of Mechanical Engineering,
The University of Texas at Austin,
Austin 204 E. Dean Keeton Street,
Stop C2200,
Austin, TX 78712
e-mail: rneptune@mail.utexas.edu

Arian Vistamehr
Brooks Rehabilitation Motion Analysis Center,
Jacksonville, FL 32216

Walking can be exceedingly complex to analyze due to highly nonlinear multibody dynamics, nonlinear relationships between muscle excitations and resulting muscle forces, dynamic coupling that allows muscles to accelerate joints and segments they do not span, and redundant muscle control. Walking requires the successful execution of a number of biomechanical functions such as providing body support, forward propulsion, and balance control, with specific muscle groups contributing to their execution. Thus, muscle injury or neurological impairment that affects muscle output can alter the successful execution of these functions and impair walking performance. The loss of balance control in particular can result in falls and subsequent injuries that lead to the loss of mobility and functional independence. Thus, it is important to assess the mechanisms used to control balance in clinical populations using reliable methods with the ultimate goal of improving rehabilitation outcomes. In this review, we highlight common clinical and laboratory-based measures used to assess balance control and their potential limitations, show how these measures have been used to analyze balance in several clinical populations, and consider the translation of specific laboratory-based measures from the research laboratory to the clinic. [DOI: 10.1115/1.4042170]

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Introduction

Human movement can be exceedingly complex to analyze due to highly nonlinear multibody dynamics, nonlinear relationships between muscle excitations and resulting muscle forces, dynamic coupling that allows muscles to accelerate joints and segments they do not span, and redundancy in muscle control [1,2]. Walking involves the execution of critical biomechanical functions such as providing body support, forward propulsion, leg swing, foot clearance/placement, and balance control that often requires synergistic activity between multiple muscle groups. A number of studies have used musculoskeletal modeling and simulation to analyze in detail individual muscle contributions to these biomechanical functions (e.g., Refs. [3–5]). From these studies, specific muscle groups have been identified as essential to the successful execution of these functions. For example, the ankle plantarflexors have been shown to provide body support and forward propulsion, accelerate the leg into swing [3–5], and provide frontal and sagittal plane balance control [6–8]. Thus, ankle muscle injury or impairment can dramatically hinder the walking ability in various populations such as those with traumatic injuries (e.g., lower-limb amputees) or impaired neural control (e.g., individuals poststroke). Of particular importance is the successful execution of maintaining dynamic balance, with the loss of balance potentially leading to falls and subsequent long-term injuries that can result in the loss of mobility and functional independence. As a result, significant research has been devoted to developing effective methods to assess balance control. In this review, we highlight common clinical and laboratory-based measures used to assess balance control and their potential limitations, show how these measures have been used to analyze balance in several clinical populations, and consider the translation of specific laboratory-based measures from the research laboratory to the clinic.

[1]Corresponding author.

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Methods to Assess Balance Performance

A number of methods have been used to evaluate balance control during human movement. These methods range from simple ordinal scale clinical balance measures to more comprehensive kinematic and kinetic-based measures derived in research laboratories.

Clinical Balance Measures. Clinical balance measures are often based on discrete score assignments while completing a series of movement tasks. The Tinetti Performance Oriented Mobility Assessment (POMA) is designed to measure balance and gait function in elderly [9]. The assessment consists of nine balance items involving sit to stand tasks and eight gait items focused on spatiotemporal gait characteristics. The Tinetti POMA measure uses a cutoff score (<20) to predict fall risk in individuals poststroke [10] and those with Parkinson disease [11]. One limitation of this measure is it has shown ceiling effects (e.g., Ref. [12]).

Berg balance scale (BBS) is used to assess functional balance and consists of 14 tasks ranging from sitting to standing and turning [13]. BBS is perhaps the most commonly used balance measure in stroke rehabilitation [14], although it is not a measure of dynamic balance, but uses a cutoff score (<42) that relates to a higher risk of falls (e.g., Ref. [15]). BBS has been shown to be an effective method for balance assessment in individuals poststroke, although some studies have observed floor and ceiling effects, and thus, combining it with other balance measures has been recommended [16]. BBS has also been shown to be a valid measure of balance in lower-limb amputees, although it has not shown promise to discriminate those with higher versus lower risk of falls [17].

Dynamic gait index (DGI) is widely used to assess dynamic balance during gait activities, which was developed by Shumway-Cook and Woollacott to evaluate functional stability and risk of falling in older adults [18]. Similar to BBS, DGI utilizes a cutoff score (<19) to indicate increased risk of falls [19,20]. In addition to older adults, DGI has been used to assess balance in individuals with vestibular dysfunction (e.g., Ref. [21]), chronic stroke (e.g., Ref. [22]), Parkinson disease (e.g., Ref. [23]), and multiple sclerosis (e.g., Ref. [24]).

Like the Tinetti POMA and BBS, DGI has shown ceiling effects in both individuals with vestibular dysfunction (e.g., Ref. [21]) and poststroke hemiparesis [25]. Moreover, in higher functioning elderly, all three measures have shown low sensitivity to responsiveness and change in scores [26]. Thus, Wrisley et al. [27] designed the functional gait assessment to address the ceiling effects. Importantly, this was done by evaluating the ceiling effects of the DGI and including more challenging tasks such as walking with a narrow base of support, walking with closed eyes and backward walking. Functional gait assessment has shown the least ceiling effect among the clinical balance scores and has been recommended for assessments in high level individuals poststroke [25]. In addition, the community balance and mobility scale was developed to assess high level deficit in balance and mobility [28]. Community balance and mobility includes tasks such as forward and backward walking, running with controlled stop, descending stairs, stepping up and dual tasking and did not show floor or ceiling effects in individuals poststroke with mild to moderate impairment [29].

In addition to potential floor and ceiling effects, an important limitation of the clinical measures is that they are global in nature and limited in their ability to provide insight into the underlying mechanisms for balance control and the biomechanical deficits linked to balance impairments. Understanding such deficits and mechanisms is needed to inform rehabilitation interventions.

Laboratory-Based Balance Measures. Laboratory-based measures can provide continuous measurements obtained using kinematic and kinetic data during walking both overground and/or on a treadmill. One commonly used measure is margin-of-stability (MoS), which is defined as the minimum distance between the base of support and the extrapolated center-of-mass (CoM) [30]. MoS is based on foot placement while accounting for body CoM position and velocity and has been used to assess dynamic balance in various populations such as in young healthy individuals in destabilizing environments (e.g., Ref. [31]), older adults while stepping to targets [32], amputees during various movement activities (e.g., Refs. [33–35]), and individuals poststroke during walking (e.g., Refs. [36] and [37]). One limitation of MoS is that it is a global measure of whole-body movement, and similar to clinical measures, does not provide insight into the biomechanical mechanisms used to control balance.

Another laboratory-based measure is whole-body angular momentum (H), which is a mechanics-based measure defined with respect to the body CoM as

$$\mathbf{H} = \sum_{i=1}^{n}[(\mathbf{r}_i^{\mathrm{COM}} - \mathbf{r}_{\mathrm{body}}^{\mathrm{COM}}) \times m_i(\mathbf{v}_i^{\mathrm{COM}} - \mathbf{v}_{\mathrm{body}}^{\mathrm{COM}}) + I_i\boldsymbol{\omega}_l]$$

where $\mathbf{r}_i^{\mathrm{COM}}$ and $\mathbf{v}_i^{\mathrm{COM}}$ are the position and velocity vectors of the ith segment's CoM, respectively. $\mathbf{r}_{\mathrm{body}}^{\mathrm{COM}}$ and $\mathbf{v}_{\mathrm{body}}^{\mathrm{COM}}$ are the position and velocity vectors of the whole-body CoM, respectively, m_i, I_i, and $\boldsymbol{\omega}_l$ are the mass, moment of inertia, and the angular velocity of each segment, respectively, and n is the number of body segments. We have performed a number of experimental and simulation studies using whole-body angular momentum to investigate the control of dynamic balance over a range of walking tasks including steady-state walking [6,7], walking at increasing speeds [38], walking with a unilateral solid ankle-foot orthosis [39], stepping on uneven terrain [40], incline/decline walking [41], and stair ascent/descent [42] and in different subject populations including amputees [38] and individuals poststroke during steady-state walking [43] and during walking adaptability tasks [44]. Others have used H to investigate balance control in younger and older

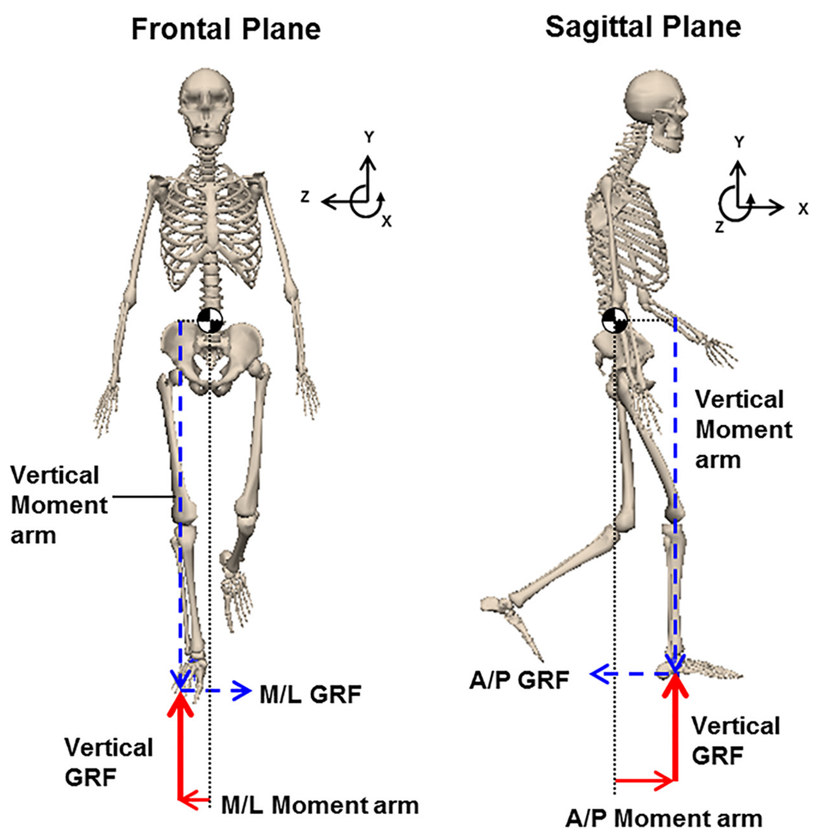

Fig. 1 The components of net external moment in the frontal and sagittal planes during single-leg stance. Whole-body CoM is shown with "◐" The GRF vectors and their corresponding moment arms appear in the same line type.

adults recovering from a trip [45], amputees using powered ankle-foot prostheses during stair ascent [46] and walking at different speeds [47], amputees during perturbed walking [48], healthy adults during multidirectional perturbed walking [49], and children with cerebral palsy walking overground [50].

The regulation of H is essential for maintaining dynamic balance during walking (e.g., Ref. [51]) and can be quantified by analyzing the time rate of change of angular momentum about the body's CoM, which is equivalent to the net external moment ($\mathbf{M}_{\text{ext}}$) (i.e., the cross-product of the moment arm vector ($\mathbf{r}$) and ground reaction force (GRF) vector) as

$$\frac{d\mathbf{H}}{dt} = \mathbf{M}_{\text{ext}}, \text{ where } \mathbf{M}_{\text{ext}} = \mathbf{r} \times \mathbf{GRF}$$

where $\mathbf{r}$ is the moment arm vector from the body CoM to the center-of pressure and **GRF** is the vector of GRFs (Fig. 1). For example, adjustments in the GRFs and foot placement in the vertical and mediolateral directions influence the net external moment in the frontal plane (Fig. 1). A higher external moment or time rate of change of H results in a higher peak-to-peak (max–min) range of H (H_R) during that time interval, which may impose a challenge to balance control.

Comparison Between Balance Measures. We previously investigated the associations between the time rate of change of frontal plane H in six regions of the gait cycle during steady-state walking with the DGI and BBS measures in individuals poststroke [43]. We found a correlation between higher clinical scores (i.e., better balance control) and lower rate of change of frontal plane angular momentum during the paretic single-leg stance. In addition, we classified the subjects as fallers or nonfallers based on their BBS and DGI scores and compared their rate of change of H between the two groups. We found that during the paretic leg single stance, fallers (based on BBS) had a significantly higher rate of change of H than nonfallers (Fig. 2). The higher rate of change of H during steady-state walking can be attributed to nonoptimal cancellation of external moment components in the fallers due to altered foot placement and/or impaired GRFs. The higher rate of change of H and corresponding H_R can be more challenging to control, particularly in impaired populations.

In a subsequent study, we built upon these findings and assessed dynamic balance using the BBS, DGI, MoS, and frontal plane H_R to determine whether these measures provided consistent assessments of balance control [52]. Correlation analyses revealed moderate associations between all measures. Overall, a higher H_R was associated with a higher MoS, wider step width, and lower BBS and DGI scores, which indicate poor balance control.

Given that balance is multidimensional, each measure can assess different constructs of dynamic balance. The advantage of clinical balance scores is that they provide a simple global assessment, which does not require the collection and processing of more complex body-segment kinematic and kinetic data. However, contrary to the laboratory-based measures, the clinical balance scores are limited in their ability to provide insight into the biomechanical mechanisms for maintaining dynamic balance or the biomechanical mechanisms leading to the loss of balance. Similarly, MoS can provide some insight into foot placement, but not the GRFs, which are generated primarily by muscle forces and are critical to the regulation of dynamic balance. Although the analysis is more complex to perform, H has a number of advantages such as it can provide a more comprehensive assessment of dynamic balance since it accounts for the motion and inertia of all the body segments, which collectively generate the whole-body angular momentum about the center-of-mass. Further, the analysis of the rate of change of H or net external moment can provide insight into the influence of foot placement and GRF generation on maintaining dynamic balance. Thus, the analysis of H reflects not only direct balance control strategies such as from muscle force generation, but also indirect methods such as counter rotation strategies using arm swing and trunk motion. Thus, the relationships between the H trajectories and corresponding GRFs and moment arms (i.e., foot placements) (Fig. 3) can be analyzed to gain insight into the biomechanical mechanisms of balance control, which reflect the whole-body response used to maintain balance. Finally, the analysis of H can be performed during specific movement tasks, in each anatomical plane independently and across a wide range of participants with no reported ceiling effects. Thus, H can provide an objective method for monitoring the biomechanical changes in balance control and assessing the effectiveness of specific balance training programs.

Muscle Contributions to Dynamic Balance

Muscles play a critical role in maintaining dynamic balance during human walking. Thus, understanding which muscles contribute to balance control has the potential to help diagnose and treat balance disorders. We previously used musculoskeletal modeling and simulation analyses to quantify individual muscle contributions to the regulation of H in the sagittal [6] and frontal [7] planes. Modeling and simulation is a powerful tool that allows one to identify the causal relationships between individual muscle excitations and the performance of specific biomechanical functions. The simulation results revealed that the regulation of H was provided by a few dominant muscle groups. In the sagittal plane, in early stance, the gluteus maximus, biarticular hamstrings, ankle

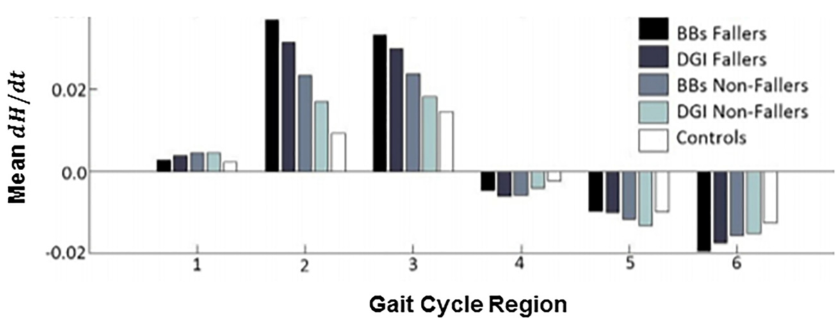

Fig. 2 The mean, time rate of change of H ($\dot{H}$) in the frontal plane during the six regions of the gait cycle during steady-state walking. Each bar depicts the mean values across the subjects grouped as fallers and nonfallers based on their BBS and DGI scores. There is a significant difference in $\dot{H}$ between the BBS fallers and nonfallers during the paretic leg single stance (regions 2 and 3). Region 1 is the first double support phase, regions 2 and 3 are the first and second halves of single-leg stance, respectively, region 4 is the second double support phase, and regions 5 and 6 are the first and second halves of swing, respectively. Figure adopted from Ref. [43].

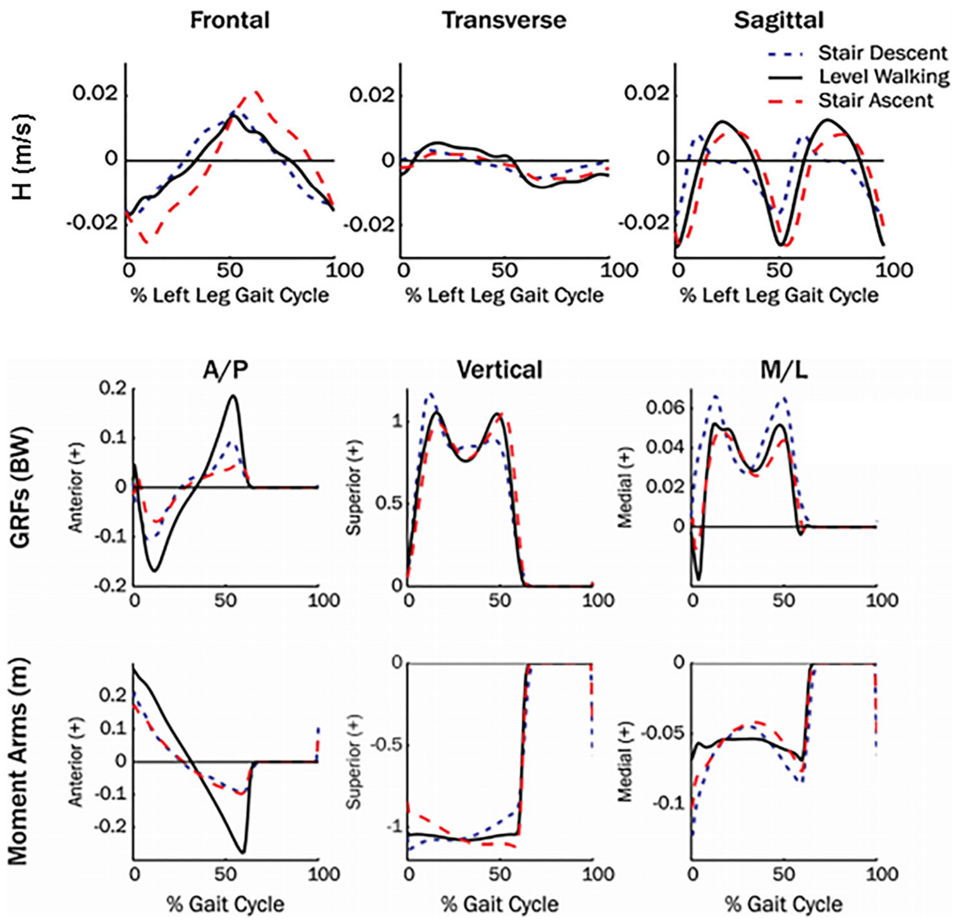

Fig. 3 Mean three-dimensional (3D) *H* trajectories, normalized by height and weight during healthy adult level walking, and stair descent and ascent. Mean GRFs and moment arms during each walking condition are shown in the anterior-posterior (A/P), vertical and mediolateral (M/L) directions. Figure adapted from Ref. [42].

dorsiflexors, and gravity contributed to the backward external moment (i.e., acted to rotate the body backward), while the soleus, gastrocnemius, and rectus femoris contributed to the forward external moment (i.e., acted to rotate the body forward). In late stance, the soleus and gastrocnemius generated angular momentum in opposite directions due to differences in their relative contributions to the horizontal and vertical ground reaction forces. The soleus generated primarily forward momentum, while the gastrocnemius generated backward momentum.

In the frontal plane, previous simulation studies have investigated mediolateral body control by analyzing muscle contributions to the linear accelerations of the whole-body center-of-mass [8,53,54] or frontal plane trunk angular accelerations [55]. We extended this work by analyzing muscle contributions to the regulation of H and found in early stance the vasti, adductor magnus, and gravity acted to rotate the body toward the contralateral leg, while the gluteus medius acted to rotate the body toward the ipsilateral leg (Fig. 4). In late stance, the gluteus medius continued to rotate the body toward the ipsilateral leg, while the soleus and gastrocnemius acted to rotate the body toward contralateral leg (Fig. 4).

An important finding in these studies was the critical role the ankle plantarflexors play in maintaining dynamic balance during walking in both the frontal and sagittal planes. Others have also shown that the plantarflexors are important in balance recovery during walking [45] and standing [56] perturbations and that individuals with a history of falls have reduced ankle plantarflexor output [57]. The critical role of the plantarflexors in maintaining dynamic balance has important implications for the diagnosis and treatment of movement and balance disorders, and also in the design and prescription of ankle-foot orthotic (AFO) devices.

What is the Role of Step Width in Balance Control?

Previous studies have suggested that wider steps are used to increase lateral stability [58–61] and some have observed that elderly fallers take narrower steps [62]. Others have associated wider steps with increased step width variability and increased instability [63] and some have observed a higher rate of falls in subjects with wider steps [64–67]. Lower-limb amputees often walk with wider steps and a more variable base of support [35,68] and are more likely to fall compared to nonamputees [69]. We found that in individuals poststroke, a wider step width was associated with a greater H_R, a higher MoS, and lower BBS and DGI scores, which indicates poor balance control [52]. Wider step widths during single limb-stance create a greater moment-arm from the body's center-of-mass to the center-of-pressure that along with the vertical GRF act to rotate the body toward the contralateral limb and increase H_R. A higher H_R most likely makes the individual more susceptible to falling when they are near their peak H and a perturbation occurs in the same direction. This presents a challenge for individuals with muscle weakness or neuromotor impairments who do not have the neuromuscular capacity to provide a timely response to counteract the perturbation through proper foot placement and/or generation of appropriate GRFs. Perturbation studies are needed to assess the relationships between H_R, neuromuscular capacity, and the ability to recover balance in various clinical populations.

MoS can provide insight into foot placement while accounting for center-of-mass position and velocity (see *Methods to Assess Balance Performance* section). Prior studies (e.g., Ref. [32]) have interpreted higher MoS in older adults as indicative of better

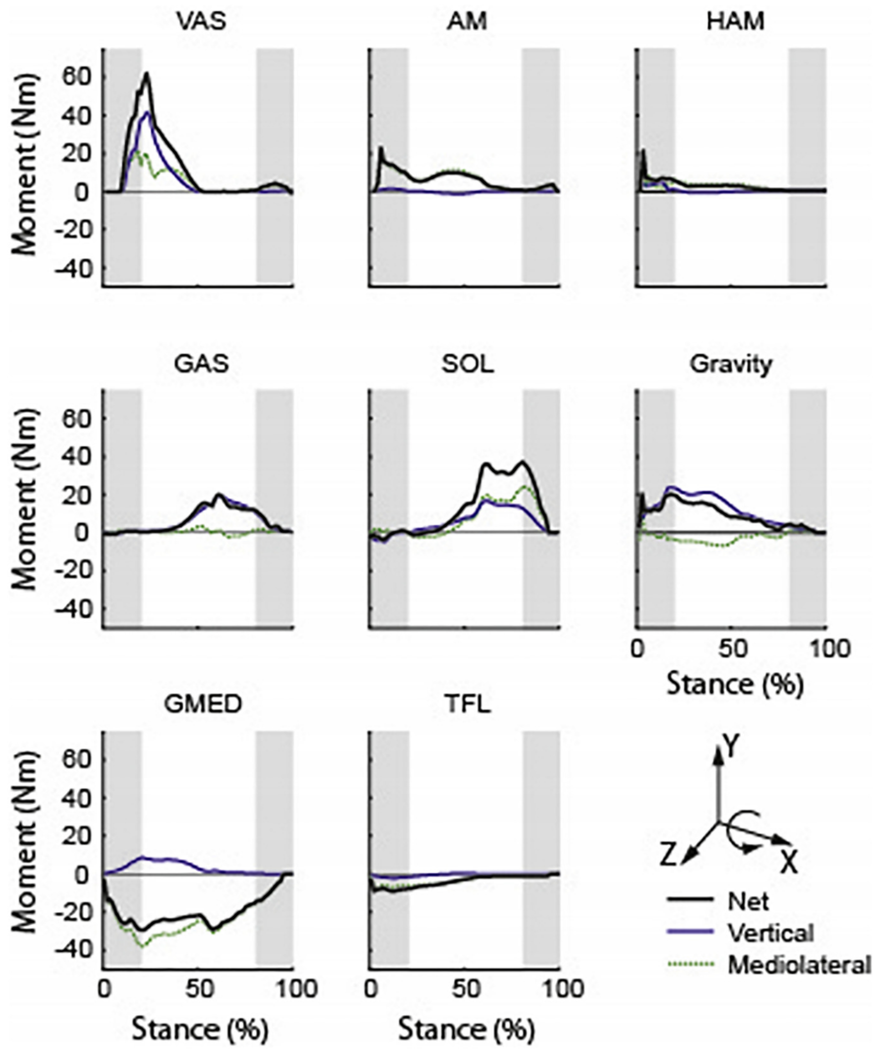

Fig. 4 Individual muscle contributions to the time rate of change of frontal plane *H* (i.e., external moment). "vertical" and "mediolateral" are contributions from the vertical and mediolateral GRFs, respectively. Positive (negative) values indicate the moment contribution from that muscle acts to rotate the body toward the contralateral (ipsilateral) leg. Shaded regions indicate the double support phases. Abbreviations: VAS, vasti; AM, adductor magnus; HAM, hamstrings; GAS, gastrocnemii; SOL, soleus; GMED, gluteus medius; and TFL, tensor fasciae latae. Figure adapted from Ref. [7].

dynamic stability compared to younger adults, and a similar interpretation of MoS was made in individuals poststroke (e.g., Ref. [70]). However, based on the correlations between larger MoS and lower BBS and DGI scores [52] and the higher rate of falls in both older adults [71] and individuals poststroke [72], an alternative interpretation is that individuals with higher MoS may be at a higher risk of falling. It is not clear if the higher MoS is adopted as a compensatory mechanism for those with poor balance control or if an increased level of MoS may paradoxically represent decreased stability during walking. When we examined MoS for each foot, we found that only the MoS corresponding to the paretic foot placement was correlated with other balance measures [52]. This highlights the importance of the paretic mediolateral foot placement in dynamic balance control, suggesting that in individuals poststroke it may be more suitable to examine the MoS for each foot individually rather than the average sum of both legs (e.g., Refs. [36] and [70]).

Previous research has shown that mediolateral foot placement requires active recruitment of the sensory-motor processes [73] and that overall frontal plane movements require greater active control than sagittal plane movements [58]. In young healthy adults, changes in the lateral foot placement were correlated with gluteus medius muscle activity [74]. Others have shown that poststroke nonfallers and healthy adults used a similar neuromechanical strategy to control their mediolateral foot placement, which was influenced by the swing phase gluteus medius activation and associated with the state of the contralateral stance limb. However, in poststroke fallers, this strategy was disrupted, especially when taking a step with the paretic leg [75].

Individuals poststroke have shown significantly lower levels of accuracy and precision in their mediolateral foot placement during walking compared to healthy controls. The lowest accuracy was observed during extreme (i.e., narrowest and widest) targeted values [76]. Others have observed that individuals poststroke had difficulty controlling their step width and foot placement variability during a gait tracking task. Specifically, the variability in their step width and paretic foot placement increased as the targeted step width decreased, highlighting task-dependent inter-limb differences in frontal plane motor control variability [77]. These studies collectively highlight the importance of mediolateral foot placement in dynamic balance control, and deficits in foot placement should be a focus in balance training programs.

Given the importance of mediolateral foot placement and step width in dynamic balance control, it would be extremely insightful to be able to measure these quantities in the clinic. In contrast to individual foot placement (i.e., moment arm) which involves the more complex calculation of body center-of-mass location, step width has been measured in the clinic using various methods. Less expensive methods involve specially designed color coded walkway grids, which register the footfalls using ink marks [78], and pressure-sensitive papers [79]. More expensive methods involve instrumented walkways registering footfalls using pressure sensors (e.g., GAITRITE) [80]. Future work is needed to further understand the role of foot placement in balance control and how simple, clinic-based methods can be used to inform treatment decisions.

Balance Control in Lower-Limb Amputees

Below-knee amputees have an increased risk of falling relative to nonamputees [69]. To begin understanding the balance control deficits in amputees, we examined H between 12 amputees and 10 nonamputees at four walking speeds ranging from 0.6 to 1.5 m/s with 0.3m/s increments [38]. In the frontal plane, H_R over the entire gait cycle was found to be greater in amputees compared to nonamputees at the first three walking speeds (Fig. 5), which was correlated with a reduced vertical GRF peak during late stance in both the sound and residual legs. In the sagittal plane, the amputee H_R in the first half of the residual leg gait cycle was significantly larger than in the nonamputees at the three highest speeds. In the second half of the gait cycle, the sagittal plane H_R was significantly smaller in amputees compared to nonamputees at all speeds. Correlation analyses suggested that the greater H_R in the first half of the amputee gait cycle is associated with reduced residual leg braking GRF peak and that the smaller H_R in the second half of the gait cycle is associated with reduced residual leg propulsive GRF peak. Thus, reducing residual leg braking appears to be an important compensatory mechanism to help regulate the sagittal plane angular momentum over the gait cycle, but the increased H in the first half of the gait cycle may lead to an increased risk of falling.

In a subsequent study, we compared H during stair ascent and descent using passive and powered lower-limb prostheses with the expectation that the powered prosthesis would eliminate the propulsion deficit and improve balance control [46]. Similar to our steady-state walking study, amputees had a larger H_R in the sagittal plane during prosthetic limb stance compared to nonamputees during stair ascent. However, there were no differences in H_R between the passive and powered prostheses in the frontal, transverse, or sagittal planes during stair ascent or descent. These results suggest that amputees have altered angular momentum trajectories during stair walking compared to able-bodied individuals, which may contribute to their increased fall risk. The results also suggest that powered prostheses provide no distinct advantage over passive prostheses in maintaining dynamic balance during stair walking. This may be due to the inability of current ankle-foot prosthetic devices to replicate the functions of both the uniarticular soleus and the biarticular gastrocnemii, which often have distinctly different biomechanical functions [81,82]. In contrast to these results, others found that amputees more effectively regulate H_R at some walking speeds when using a powered prosthesis compared with passive-elastic prostheses [47]. This finding is likely due to forward propulsion generation being dominated by the uniarticular soleus, which can be effectively replaced by a powered ankle-foot prosthesis.

We further analyzed the contributions of a passive prosthesis and individual muscles to balance control during amputee and nonamputee stair ascent [83] and found that the passive prosthesis replicated the role of nonamputee plantarflexors in the sagittal plane but caused a larger change in angular momentum in the transverse plane. In the frontal plane, nonamputee plantarflexors contributed minimally, while the prosthesis was a critical contributor to angular momentum that acted to rotate the body toward the contralateral leg. This resulted in altered muscle contributions from the vasti, hamstrings, and hip abductors. These results suggest that improved prostheses with reduced contributions to transverse and frontal plane angular momentum could improve dynamic balance during amputee stair ascent and minimize necessary muscle compensations.

Most studies have analyzed balance control during steady-state conditions. However, perturbations frequently occur during walking and compromise balance. Studies have shown that the

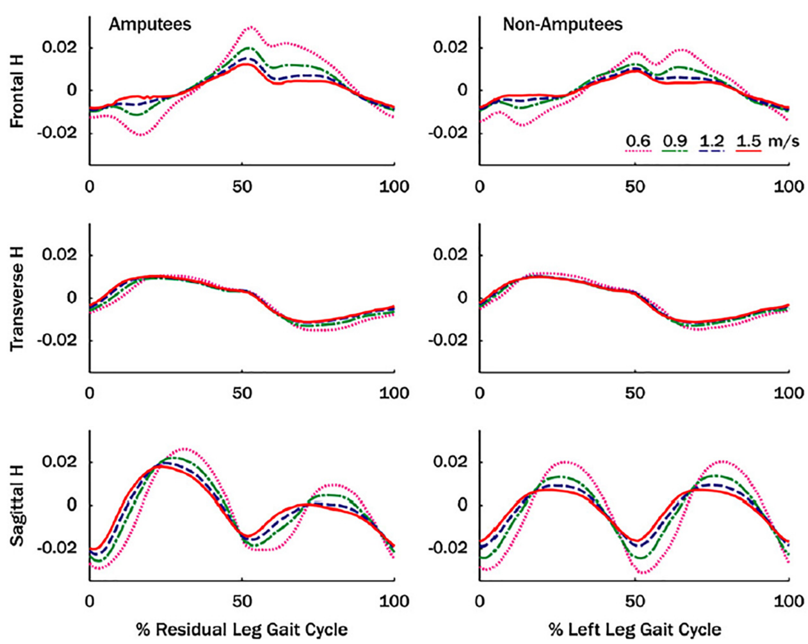

Fig. 5 Mean normalized three-dimensional angular momentum (*H*) for amputee and nonamputee subjects over the gait cycle. Angular momentum was normalized by body mass, body height, and walking speed. Note, in this study, *H* was normalized by walking speed which can influence the magnitude of *H* across the walking speeds. Figure adapted from Ref. [38].

majority of falls occur due to trips or unexpected perturbations [84]. Perturbations are particularly challenging for lower limb amputees who lack active ankle control to help recover their balance. Thus, understanding the neuromuscular balance recovery mechanisms used by amputees to recover from such perturbations could provide insight into developing interventions aimed at decreasing their risk for falls and injuries. Sheehan et al. [48] assessed the effects of surface perturbations on balance control in able-bodied controls and individuals with unilateral transtibial amputation using mediolateral platform oscillations. Amputees and nonamputees walked at a fixed speed with no perturbations and continuous, pseudo-random, mediolateral platform oscillations. Amputees were significantly more affected by the perturbations and had a greater H_R in the frontal plane. They noted that their findings support the use of angular momentum in relation to dynamic stability and that increased H_R is associated with greater fall risk.

Segal and Klute [85] used a novel mediolateral foot perturbation protocol that enabled them to study the effect of medial and lateral step width disturbances on balance control in both amputees ($n = 10$) and nonamputees ($n = 12$). They used a pneumatic device attached to the foot to release a medial or lateral burst of air just before heel strike that imposed a repeatable medial or lateral disturbance in foot placement. They found amputees required five steps to return to undisturbed step width after a prosthetic limb medial disturbance versus two steps for the sound limb and for nonamputees. Following a lateral disturbance, amputees returned to their undisturbed step width within three steps, which was similar to the sound and nonamputee limbs. Thus, for amputees, a medial perturbation was much more challenging than a lateral perturbation in terms of the number of steps needed to recover their balance.

In a follow-up study, we analyzed the same dataset to further understand the balance recovery mechanisms used by lower limb amputees in response to the perturbations by examining changes to frontal plane H and hip joint work [86]. The lateral perturbations of the residual, sound, and nonamputee limbs resulted in a reduced H_R and an increased positive frontal plane hip work in the first half of single limb support. Medial perturbations for all limbs resulted in increased H_R and decreased positive frontal plane hip work, also in the first half of single limb support. These results further support the important role hip strategies play in balance control. Thus, rehabilitation interventions that focus on hip muscles that regulate mediolateral balance, particularly the hip abductors, and the use of prostheses with active ankle control, may reduce the risk of falls in lower-limb amputees.

Another common gait disturbance that can lead to the loss of balance and falls is stepping on unpredictable terrain. We previously identified the biomechanical response to a step on coronally uneven and unpredictable terrain [40]. Able-bodied subjects traversed a walkway with a middle step that was blinded to participants, and positioned either 15 deg inverted, 15 deg everted or flush. The analysis of H_R in the frontal plane showed that it increased during blinded eversion and decreased during blinded inversion (Fig. 6). In the frontal plane, the analysis of external moments applied to the body about the center-of-mass by the disturbed and recovery legs suggested that the disturbed leg contributed more to differences in H_R, and thus, to balance recovery. During the disturbed step, distinct differences between blinded inversion and eversion in the frontal plane moments of the hip and ankle suggested that the hip and ankle joint moment strategies were important for adapting to the terrain angle. Thus, amputees would most likely have difficulty adapting to such balance perturbations without an active ankle strategy.

These studies of individuals with below-knee amputations and balance perturbations have highlighted the usefulness of H in gaining insight into balance control mechanisms and provided the basis for targeted rehabilitation programs to strengthen specific muscle groups and help improve dynamic balance and subsequently reduce the risk of falls and injuries.

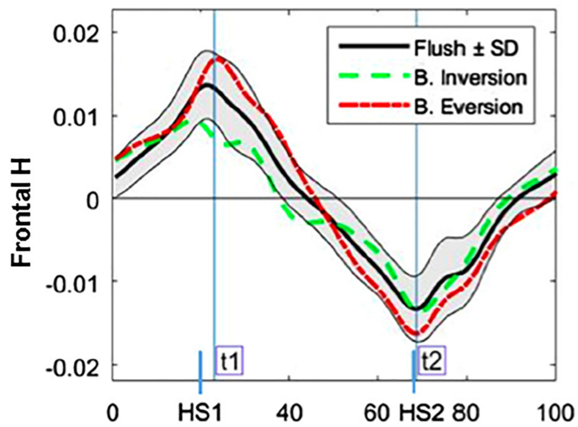

Fig. 6 Mean normalized *H* in the frontal plane for each perturbation condition. The everted perturbations resulted in greater H_R. Figure adapted from Ref. [40].

Balance in Individuals Poststroke

More than 50% of stroke survivors experience falls within one year poststroke (e.g., Ref. [87]), which can lead to physical injuries and long-term disabilities. Although 85% of individuals poststroke regain some level of steady-state walking function, over one-third of these individuals reportedly do not walk in the community (e.g., Ref. [88]). Community walking involves the performance of adaptability tasks such as obstacle negotiation, stepping up a curb, and changing the walking direction [89]. Successful performance of these tasks requires precise balance control, which is a major challenge poststroke.

During steady-state walking, Nott et al. [43] showed that healthy adults demonstrated timely regulation of frontal plane H during the first half of single-leg stance, with the level of regulation depending on the initial magnitude. In contrast, individuals poststroke who poorly regulated their frontal plane H during initial paretic leg single stance exhibited lower DGI and BBS scores and some were categorized as fallers.

To gain insight into balance control impairments during community ambulation, we have recently studied the regulation of H in all three anatomical planes in 15 individuals poststroke and 10 healthy adults during a variety of walking tasks including steady-state self-selected and fastest-comfortable walking, backward walking, obstacle negotiation, and stepping up a box (Step) [44]. We observed that individuals poststroke had significant deficits regulating their H in the frontal plane, manifested in significantly higher H_R than the healthy adults (Fig. 7). Further, the obstacle negotiation task was associated with a higher H_R compared to the

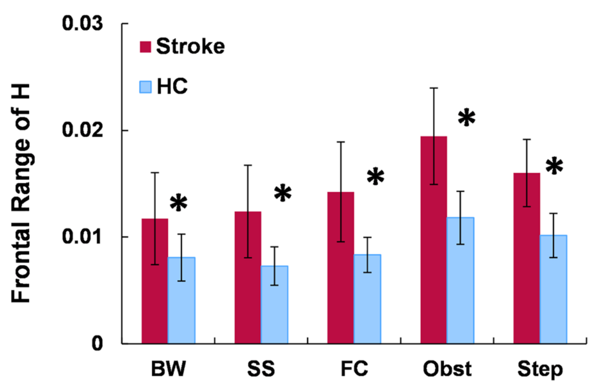

Fig. 7 Mean normalized range of *H* (H_R) in the frontal plane during backward, self-selected, and fastest-comfortable forward walking as well as obstacle negotiation and stepping up (Step). A significant difference between poststroke and healthy controls is indicated with *($P < 0.05$).

rest of the tasks and imposed a higher demand for balance control particularly in individuals poststroke. In addition, in contrast to healthy adults who had higher soleus activation levels during the adaptability tasks compared to steady-state walking, minimal changes in the soleus activation levels were evident in individuals poststroke [44]. Thus, the inability to recruit the plantarflexors, which are critical in controlling dynamic balance in the frontal plane (see *Muscle contributions to Dynamic Balance* section), may compromise dynamic balance during these more challenging adaptability tasks.

Influence of Ankle-Foot Orthoses on Balance Control

Ankle-foot orthoses are widely prescribed to improve walking ability for those with various neurological deficits by assisting with foot clearance during swing while stabilizing the ankle during stance and keeping it in a near neutral position. As a result, ankle motion and plantarflexor function during stance is limited and may hinder the ability of those with volitional plantarflexor activity to regulate angular momentum in response to dynamic perturbations, and therefore, compromise their dynamic balance.

We examined the influence of a common clinically prescribed solid unilateral polypropylene AFO on dynamic balance in healthy adults during steady-state slow (0.6 m/s) and moderate (1.2 m/s) speeds, and during accelerated (0–1.8 m/s at 0.06 m/s^2) and decelerated (1.8–0m/s at -0.06 m/s^2) walking [39]. We again used H_R to quantify dynamic balance. We found AFO use resulted in a greater H_R in both the frontal and sagittal planes (Fig. 8), which were correlated with the reduced peak hip abduction and reduced ankle plantarflexor moments, respectively. In addition, walking with the AFO decreased body forward propulsion and ankle power generation (Fig. 9). These results suggested that AFOs may hinder the successful execution of important biomechanical subtasks and ankle function in healthy adults. Clearly, for those with ankle eversion or low plantarflexor activity, AFO prescription is a useful component of their rehabilitation to provide needed foot clearance and stability. However, for those that have volitional plantarflexor activity, AFOs may limit the successful execution of important mobility subtasks.

Considering that improving the execution of walking subtasks such as balance control is an important element in rehabilitation, future research is needed to assess the tradeoffs of AFO use. For instance, in addition to examining the mechanical influence of AFOs on walking biomechanics, it is critical to identify the consequences of AFO use on muscle activity and neural plasticity during rehabilitation. We expect that AFOs will hinder the contribution of the plantarflexors and potentially require compensatory actions by other muscles to achieve balance control in individuals poststroke. This is especially critical during the acute recovery phase when neural plasticity is highest and poor muscle coordination patterns can be learned. Thus, future work is needed to critically assess the ramifications of long-term AFO use on rehabilitation outcomes.

Translation of Laboratory-Based Measures of *H* to the Clinic

One limitation of whole-body angular momentum is that the acquisition and analysis of H requires elaborate and expensive motion capture equipment that is not readily available in the clinic. Eventually, technologies such as inertial measurement unit sensors may allow for such assessments in the clinic, but currently they are limited in their capabilities. An important use of H analyses is that they can be used to gain insight into the effectiveness of common or new interventions that are considered in the clinic. For example, the analysis of H has been used to assess the effectiveness of a locomotor training program on balance control during steady-state walking in individuals poststroke [90]. Further, this analysis identified the underlying mechanisms (i.e., foot placement and GRF modifications) used from pre- to posttraining that were associated with observed improvements in dynamic balance. In addition, relationships between improvements in H post-intervention and simple biomechanical markers pretraining (e.g., walking speed) were identified to distinguish those who are likely to show improved balance control from the training program [90].

Aside from assessing the effectiveness of locomotor training interventions and specific balance training programs, the analysis of H can also be used to objectively identify movement tasks that challenge individuals' balance control and may provide a basis for improving their mobility in the community. For instance, we have used the analysis of H during a variety of walking adaptability tasks and identified that individuals poststroke were most challenged in controlling their angular momentum during obstacle negotiation [44]. The most pronounced deficits in controlling H occurred in the mediolateral direction and during single-limb-stance when the trailing limb was behind the obstacle. These deficits were associated with lack of soleus activation and a wider separation between the body center-of-mass and the paretic foot

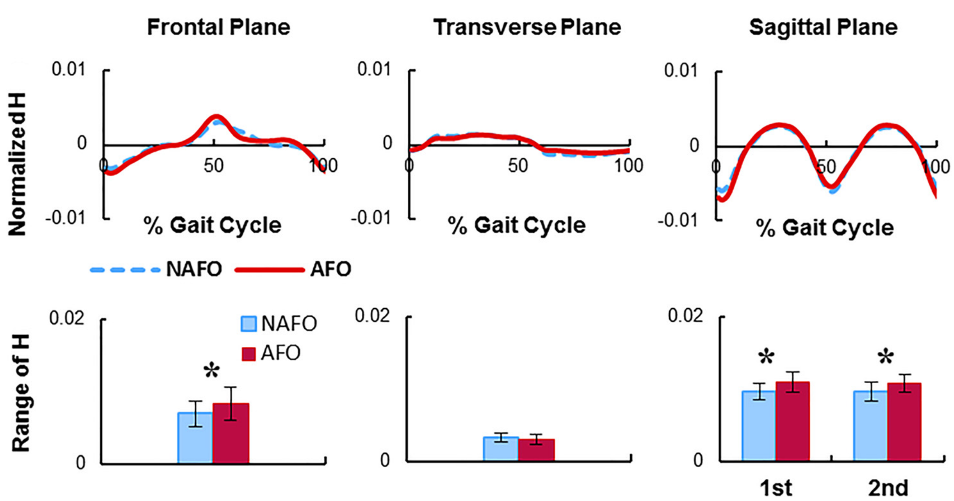

Fig. 8 Normalized, mean three-dimensional whole-body angular momentum (*H*) during steady-state walking with (AFO) and without (NAFO) a solid unilateral AFO. Figures are in the AFO leg reference frame. The H_R mean (SD) is shown in the bottom row. A significant difference between AFO and NAFO is indicated with "*" ($P < 0.05$). Figure adapted from Ref. [39].

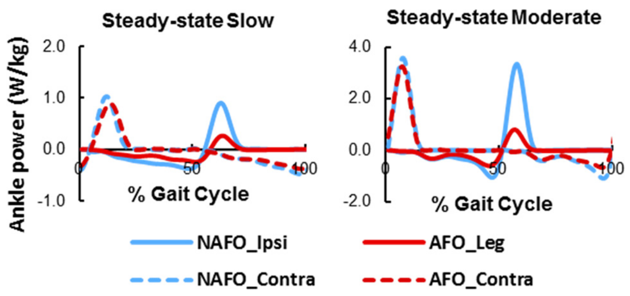

Fig. 9 Mean normalized ankle power when walking with (AFO) and without (NAFO) a solid unilateral AFO at slow (0.6 m/s) and moderate (1.2 m/s) speeds. Figure adapted from Ref. [39].

center-of-pressure (i.e., deficits in weight shifting toward the paretic limb). Thus, practicing obstacle negotiation tasks in the clinic with an emphasis on the trailing limb single-stance phase may be an effective way to improve community ambulation poststroke.

Summary

Maintaining dynamic balance during human movement is critical to preventing falls and injuries that can lead to loss of mobility and functional independence. Various clinical and laboratory-based measures can be used to assess balance control with each having their own strengths and weaknesses. We have found the analysis of whole-body angular momentum to be a powerful tool to gain insight into the underlying mechanisms for maintaining dynamic balance in healthy adults and individuals with mobility impairments. A challenge for future work is to benchmark whole-body angular momentum against fall rates and determine a threshold for the range of H that is associated with falls in specific patient populations. Another avenue for future work is to identify associations between H and other measures that are readily available in the clinic (e.g., walking speed, spatiotemporal gait characteristics, or other simple clinical measures) and investigate if these measures can help predict the effectiveness of therapeutic interventions.

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