Muscle Contributions to Balance Control During Amputee and Nonamputee Stair Ascent

Nicole G. Harper[1]
Walker Department of Mechanical Engineering,
The University of Texas at Austin,
204 E. Dean Keeton Street, Stop C2200,
Austin, TX 78712
e-mail: nicole.harper@utexas.edu

Jason M. Wilken
Extremity Trauma and Amputation
Center of Excellence,
Center for the Intrepid,
Brooke Army Medical Center,
Ft. Sam Houston, TX 78234;
Department of Physical Therapy and
Rehabilitation Science,
The University of Iowa,
1-252 Medical Education Building,
Iowa City, IA 52240
e-mail: jason-wilken@uiowa.edu

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

Dynamic balance is controlled by lower-limb muscles and is more difficult to maintain during stair ascent compared to level walking. As a result, individuals with lower-limb amputations often have difficulty ascending stairs and are more susceptible to falls. The purpose of this study was to identify the biomechanical mechanisms used by individuals with and without amputation to control dynamic balance during stair ascent. Three-dimensional muscle-actuated forward dynamics simulations of amputee and nonamputee stair ascent were developed and contributions of individual muscles, the passive prosthesis, and gravity to the time rate of change of angular momentum were determined. The prosthesis replicated the role of nonamputee plantarflexors in the sagittal plane by contributing to forward angular momentum. The prosthesis largely replicated the role of nonamputee plantarflexors in the transverse plane but resulted in a greater change of angular momentum. In the frontal plane, the prosthesis and nonamputee plantarflexors contributed oppositely during the first half of stance while during the second half of stance, the prosthesis contributed to a much smaller extent. This resulted in altered contributions from the intact leg plantarflexors, vastii and hamstrings, and the intact and residual leg hip abductors. Therefore, prosthetic devices with altered contributions to frontal-plane angular momentum could improve balance control during amputee stair ascent and minimize necessary muscle compensations. In addition, targeted training could improve the force production magnitude and timing of muscles that regulate angular momentum to improve balance control. [DOI: 10.1115/1.4047387]

1 Introduction

Stair ascent is an activity of daily living critical for maintaining independence in the home and community. However, previous studies have shown that dynamic balance is more difficult to maintain during stair ascent than level walking [1], particularly in the frontal plane [2,3]. As a result, populations with balance deficits often have difficulty ascending stairs and are at an increased risk for falls (e.g., Ref. [4]). Previous studies have attempted to mitigate fall risk using external supports such as handrails [5,6] and canes [7,8], but understanding and improving an individual's ability to control dynamic balance independent of assistive devices is critical.

Whole-body angular momentum has been identified as a valuable method for investigating the biomechanical mechanisms underlying balance control (for review, see Ref. [9]). Regulation of whole-body angular momentum has been identified as a critical aspect of maintaining dynamic balance during level walking [10] and has been used as the foundation of control algorithms for bipedal robots (e.g., Refs. [11–13]). Several studies have also suggested that controlling angular momentum is important in preventing falls during level walking [14] and in recovering from a trip [15–17], an unexpected step [18] or a step on uneven terrain [19,20]. In addition, regulation of angular momentum has been shown to be a promising indicator of balance in impaired populations [21–24].

Whole-body angular momentum, which is a measure of the body's rotation about its center-of-mass (COM), is often quantified by its time derivative (i.e., the time rate of change of angular momentum), which is equal to the external moment about the body COM. The external moment is a function of the distance between the body COM and center-of-pressure as well as the ground reaction forces (GRFs). As a result, angular momentum can be modulated through changes in the GRFs and external moment arms (e.g., foot placement). In human gait, muscles are the primary accelerators of the body segments and contributors to the GRFs and are therefore the primary regulators of whole-body angular momentum. Thus, to design effective rehabilitation programs and enhanced assistive devices aimed at improving dynamic balance, it is critical to understand how individual muscles contribute to the regulation of angular momentum in impaired and unimpaired populations.

One group of individuals who are particularly susceptible to falls is lower-limb amputees [25]. Recent studies analyzing whole-body angular momentum during level walking [24,26], perturbed walking [27], sloped walking [28], and running [29] observed an increased range of angular momentum in amputees compared to nonamputees, which suggests a decrease in dynamic balance control. In addition, increased variability in angular momentum was observed on uneven terrain [19] in amputees compared to nonamputees. Others have shown that walking with a powered prosthesis, compared to a passive energy storage and return prosthesis, diminished sagittal-plane range of angular momentum in level walking [26] and during the first half of the gait cycle in sloped walking [28,30]. This suggests that regulation of whole-body angular momentum, and thus dynamic balance, has the potential to be improved with enhanced prosthesis designs that more fully replicate the role of the plantarflexors. In stair ascent, two recent studies found that regulation of angular momentum in both amputees [2] and nonamputees [3] was significantly altered compared to level walking. Furthermore, amputees exhibited an increased range of sagittal-plane angular momentum compared to nonamputees, which suggests a decrease in balance control. Amputees also exhibited an increased range of trunk angular momentum in the frontal and transverse planes during stair ascent [31]. However, despite this increased understanding of balance control, it remains unclear how amputees regulate whole-body angular momentum during stair ascent.

Muscle-actuated forward dynamics simulations have been used successfully to identify the contributions of individual muscles to

[1]Corresponding author.
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the time rate of change of angular momentum during nonamputee level walking [32,33]. The purpose of this study was to similarly use muscle-actuated forward dynamics simulations to identify the contributions of individual muscles, the passive prosthesis, and gravity to whole-body angular momentum in the frontal, transverse, and sagittal planes to gain insight into the different biomechanical mechanisms used by individuals with and without amputation to regulate angular momentum and control dynamic balance during stair ascent. Understanding which muscles are primarily responsible for regulating angular momentum and identifying the differences between individuals with and without amputation has important implications for the design of improved prostheses and targeted training programs aimed at improving balance control during stair ascent.

2 Methods

2.1 Musculoskeletal Model and Dynamic Optimization. Previously described three-dimensional muscle-actuated forward dynamics simulations of amputee [34] and nonamputee [35] stair ascent were developed using SIMM/Dynamics Pipeline (MusculoGraphics, Inc., Santa Rosa, CA) and consisted of rigid body segments representing the head, arms, and trunk (HAT), pelvis, and two legs, each consisting of a thigh, shank, patella, talus, calcaneus, and toes. Foot-ground contact was modeled using 31 visco-elastic elements with Coulomb friction distributed over the calcaneus and toes [36] or the prosthetic foot. The height of the ground was modified to represent the surface of the stairs (Rise/Run: 0.1778 m/0.2794 m). The forward dynamics equations of motion were generated using SD/FAST (PTC, Needham, MA).

In the unilateral transtibial amputee simulation, the altered mass and inertia of the residual limb were modeled by reducing the mass of the residual shank by 50% (for additional information, see Supplemental Material on the ASME Digital Collection) and shifting the shank's COM proximally to be 25% of the knee-to-ankle distance below the knee [37]. The prosthetic foot and intact foot were modeled using similar mass and inertia characteristics (for additional information, see Supplemental Material on the ASME Digital Collection). In addition, the muscles spanning the residual ankle joint were removed (see Table 1). The passive prosthesis was modeled by fitting the average experimental amputee ankle moment data (see Sec. 2.3 for experimental data collection methodology) with a second-order torsional spring with damping, which was a function of ankle angle and angular velocity. This prosthesis torque was then applied to the residual ankle joint as a passive torque.

The nonamputee model was driven by 38 Hill-type musculotendon actuators per leg, grouped into 15 muscle groups for analysis based on similar anatomical and functional classification (see Table 1). The amputee model was driven by 69 muscles (38 on the intact leg and 31 on the residual leg), grouped into 15 muscle groups in the intact leg and 12 muscle groups in the residual leg (see Table 1).

Forward dynamics simulations of nonamputee and amputee stair ascent were generated over 100% (right foot-strike to right foot-strike) and 120% (intact foot-strike to the second residual toe-off) of the gait cycle, respectively. These durations were chosen for the purpose of using these simulations in our previous analyses that studied both stance and swing phases [34,35] and to fully capture both intact and residual stance in amputee stair ascent [34]. A simulated annealing optimization algorithm [38] was used to identify muscle excitation patterns that minimized the differences between experimental and simulated kinematics and GRFs (see Sec. 2.3 for experimental data collection methodology) in addition to minimizing muscle stress to reduce unnecessary muscle cocontraction.

2.2 Simulation Analyses. To identify the contributions of individual muscles, the passive prosthesis, and gravity to dynamic balance, their contributions to the time rate of change of whole-body angular momentum ($\dot{\overline{H}}$) [32,33] were determined using the following equation:

$$\dot{\overline{H}} = \overline{r} \times \overline{F}_{\mathrm{GRF}} \quad (1)$$

where $\overline{r}$ is the moment arm vector from the center-of-pressure on each foot to the body's COM, $\overline{F}_{\mathrm{GRF}}$ is the vector of each muscle, the prosthesis, and gravity's contributions to the ground reaction force (GRF), and $\overline{r} \times \overline{F}_{\mathrm{GRF}}$ is the vector of external moments (frontal, transverse, and sagittal planes) generated about the body's COM by each muscle, the prosthesis, and gravity (Fig. 1). These contributions were determined using a previously described GRF decomposition technique [39]. The net contribution of each muscle group, the prosthesis, and gravity to whole-body angular momentum in each plane was then determined by integrating $\dot{\overline{H}}$ over the first (weight acceptance through pull-up; see Fig. 1 in Refs. [34] and [35]) and second (forward continuance through push-up; see Fig. 1 in Refs. [34] and [35]) halves of stance.

Table 1 Muscles included in the musculoskeletal model and their corresponding analysis groups in the nonamputee, intact, and residual legs. The muscles labeled as "REMOVED" have been removed from the residual leg.

	Analysis groups		
Muscles	Nonamputee leg	Intact leg	Residual leg
Iliaucs Psoas	IL	IL	IL
Adductor longus Adductor brevis Pectineus Quadratus femoris	AL	AL	AL
Superior adductor magnus Middle adductor magnus Inferior adductor magnus	AM	AM	AM
Sartorius	SAR	SAR	SAR
Rectus femoris	RF	RF	RF
Vastus medialis Vastus lateralis Vastus intermedius	VAS	VAS	VAS
Anterior gluteus medius Middle gluteus medius Anterior gluteus minimus Middle glutces minimus	GMEDA	GMEDA	GMEDA
Posterior gluteus medius Posterior gluteus minimus Piriformis Gemellus	GMEDP	GMEDP	GMEDP
Tensor fasciae latae	TFL	TFL	TFL
Superior gluteus maximum Middle gluteus maximum Inferior gluteus maximus	GMAX	GMAX	GMAX
Semitendinosus Semimembranosus Gracilis Biceps femoris long head	HAM	HAM	HAM
Biceps femoris short head	BFSH	BFSH	BFSH
Medial gastrocnemius Lateral gastrocnemius	GAS	GAS	REMOVED
Soleus Tibialis posterior Flexor digitorum longus	SOL	SOL	REMOVED
Tibialis anterior Extensor digitorum longus	TA	TA	REMOVED

2.3 Experimental Tracking Data. Previously collected three-dimensional kinematics and GRFs from 27 individuals without amputation and ten individuals with traumatic unilateral transtibial amputations using a passive energy storage and return prosthesis were used to generate these simulations. Each subject ascended a 16-step instrumented staircase (2 forceplates, 1200 Hz: AMTI, Inc., Watertown, MA) step-over-step at a fixed cadence of 80 steps per minute while a 26-camera optoelectronic motion capture system (120 Hz, Motion Analysis Corp., Santa Rosa, CA) collected three-dimensional whole-body kinematics. For five complete gait cycles for each leg, GRFs (normalized by subject body weight) and joint kinematics were time-normalized to 100% of the gait cycle and averaged across gait cycles and subjects for each leg. For additional details on the experimental protocol for the individuals with and without amputation, see Refs. [34] and [35], respectively.

3 Results

Each muscle in addition to the prosthesis and gravity contributed to the time rate of change of whole-body angular momentum. The time-trajectories of these contributions to the external moments are presented in greater detail in the Supplemental Material (Figs. S1–S9 available in the Supplemental Material on the ASME Digital Collection) while the primary contributors to whole-body angular momentum during the first and second halves of nonamputee (right leg of nonamputee simulation), intact (right leg of amputee simulation), and residual (left leg of amputee simulation) stance are discussed below. See Table 1 for all muscle group abbreviations.

3.1 Frontal-Plane Angular Momentum. Throughout nonamputee, intact, and residual stance, the hip abductors (GMEDA, GMEDP) were the primary contributors to angular momentum that acted to rotate the body toward the ipsilateral leg (ipsilateral angular momentum; Fig. 2). Additional contributions were provided by RF and TFL. During the first half of nonamputee, intact, and residual stance, VAS was the primary contributor to angular momentum that acted to rotate the body toward the contralateral leg (contralateral angular momentum), with additional contributions from gravity, HAM, and the prosthesis during residual stance and plantarflexors (SOL and GAS) during intact stance (Fig. 2). During the second half of nonamputee stance, HAM was the primary contributor to contralateral angular momentum, with additional contributions from AL and the plantarflexors (SOL, GAS) (Fig. 2). During the second half of residual stance, gravity and HAM were the primary contributors to contralateral angular momentum while the plantarflexors (SOL, GAS) and hip adductors (AM, AL) were the primary contributors during the second half of intact stance. The prosthesis contributed minimally to contralateral angular momentum during the second half of both intact and residual stance (Fig. 2).

3.2 Transverse-Plane Angular Momentum. In the transverse plane, contributions were an order of magnitude smaller than the contributions in the frontal and sagittal planes. Therefore, for brevity the results and discussion for transverse-plane angular momentum are presented in the Supplemental Material (Fig. S10 and text available in the Supplemental Materials on the ASME Digital Collection). Briefly, the prosthesis was able to largely replicate the role of the nonamputee plantarflexors in the transverse plane but caused a greater change in transverse-plane angular momentum which necessitated additional muscle compensations.

3.3 Sagittal-Plane Angular Momentum. During the first half of nonamputee stance, the plantarflexors (particularly SOL) and RF contributed to forward (negative) angular momentum while GMAX contributed to backward (positive) angular momentum (Fig. 3). The primary contributors during the first half of residual and intact stance were largely similar with the prosthesis or plantarflexors contributing to forward angular momentum and

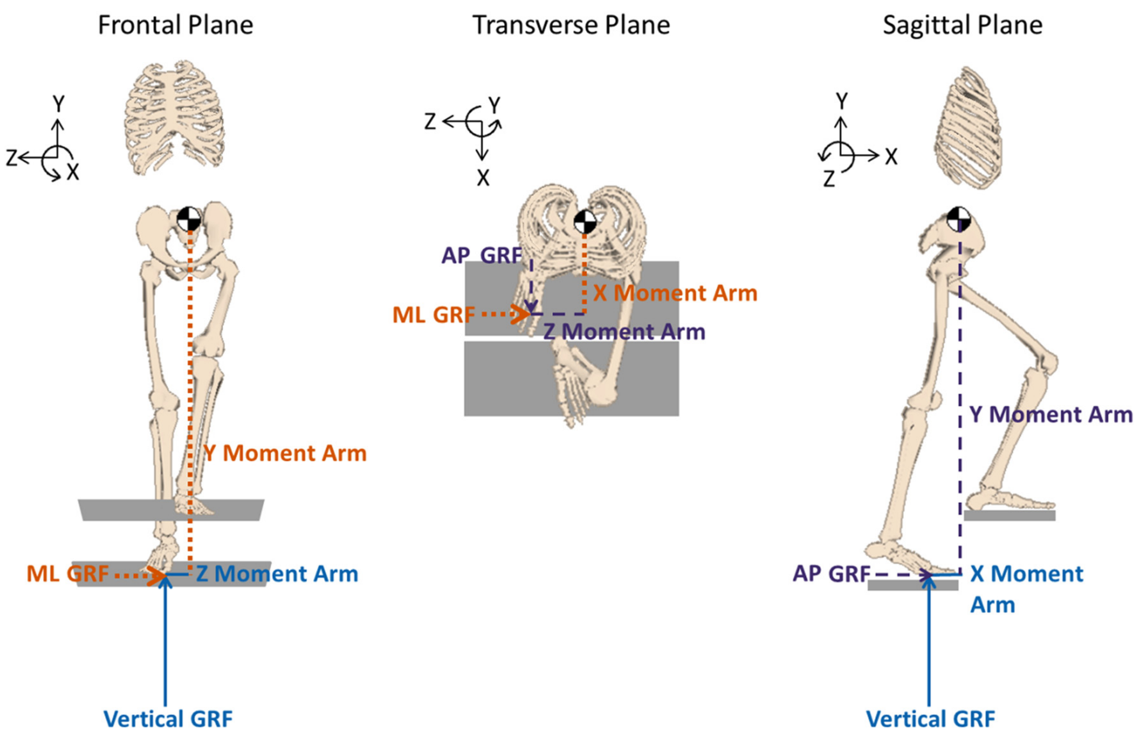

Fig. 1 The time rate of change of whole-body angular momentum (external moment about the center-of-mass (COM)) in the frontal, transverse, and sagittal planes, computed as the cross product of the external moment arms and ground reaction force (GRF) vectors (anteroposterior (AP), vertical, and mediolateral (ML)). The external moments in the frontal, transverse, and sagittal planes were defined about the *X*, *Y*, and *Z* axes, respectively. Note that for clarity, only the external moments generated by the right leg (nonamputee leg, intact leg) during stair ascent are depicted.

GMAX contributing to backward angular momentum, although HAM was a larger contributor during residual stance (Fig. 3). In addition, during the first half of stance gravity was a primary contributor to backward angular momentum in all three legs (Fig. 3). During the second half of nonamputee stance, SOL, RF, and VAS contributed to forward angular momentum while HAM contributed to backward angular momentum (Fig. 3). Similarly, during the second half of intact stance, RF, IL, and SOL contributed to forward angular momentum and HAM contributed to backward angular momentum (Fig. 3). During the second half of residual stance, the prosthesis and RF contributed to forward angular momentum, while both VAS and HAM contributed to backward angular momentum (Fig. 3).

4 Discussion

Dynamic balance is more difficult to maintain during stair ascent compared to level walking [1,40] and as a result, individuals with balance deficits such as individuals with lower-limb amputations often have difficulty ascending stairs and are more susceptible to falls [25]. During human locomotion, dynamic balance is largely regulated by controlling changes in angular momentum (e.g., Ref. [10]). Therefore, the purpose of this study was to determine the differences in the biomechanical mechanisms used by individuals with and without amputation to regulate angular momentum and control dynamic balance during stair ascent. This was accomplished by identifying the contributions of individual muscles, the prosthesis, and gravity to whole-body angular momentum.

In the sagittal plane, the prosthesis was able to largely replicate the function of the nonamputee plantarflexors during stair ascent by contributing to forward angular momentum throughout residual limb stance. The intact plantarflexors also contributed to forward angular momentum throughout intact limb stance and RF contributed to forward angular momentum across stance in all

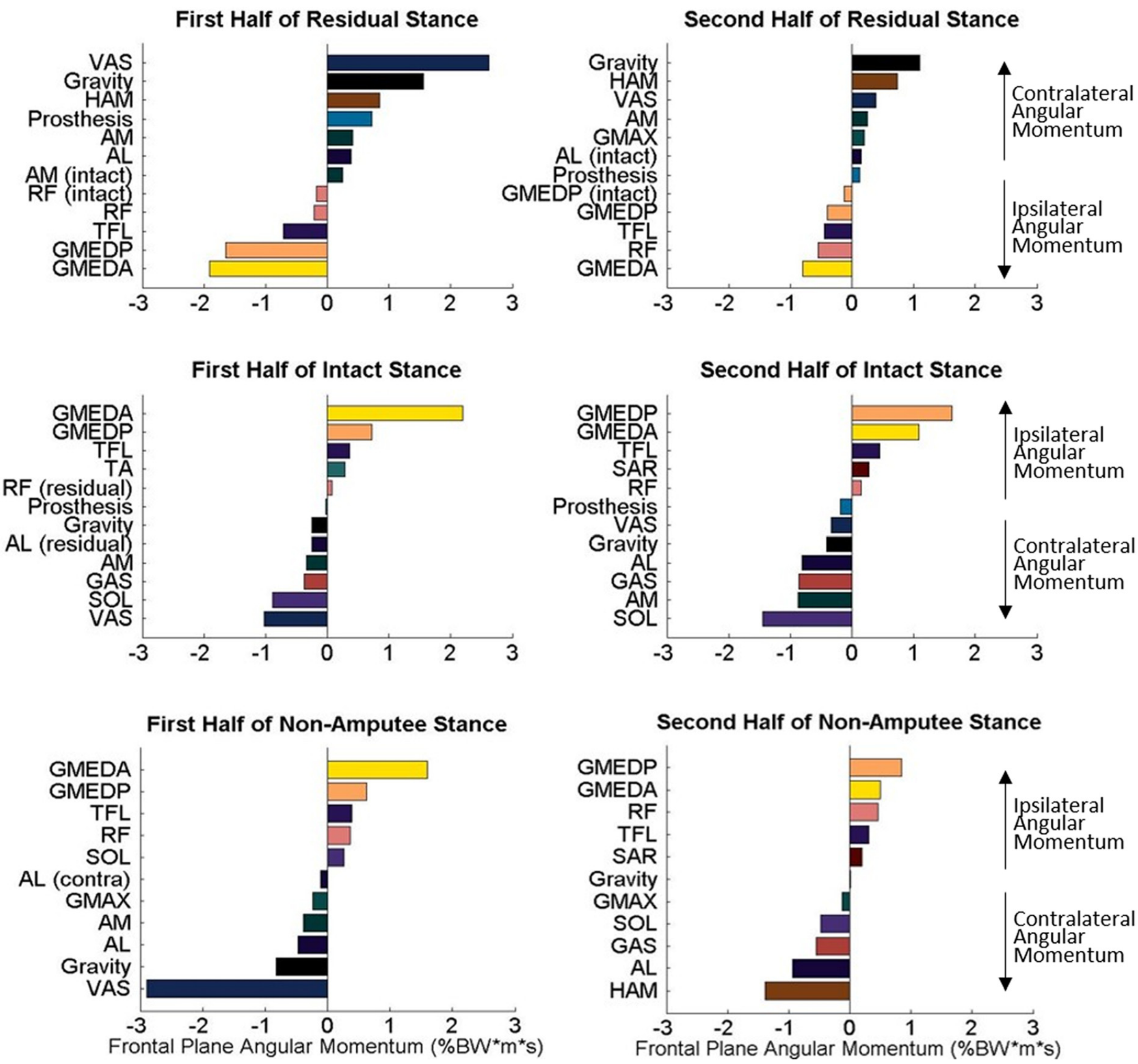

Fig. 2 Primary positive and negative muscle contributions in addition to the contributions from the prosthesis and gravity to whole-body angular momentum in the frontal plane during the first and second halves of residual, intact, and nonamputee stance during stair ascent. In residual stance (left leg of the amputee simulation), positive (negative) contributions indicate angular momentum that acts to rotate the body toward the contralateral (ipsilateral) leg. In intact and nonamputee stance (right leg of the amputee and nonamputee simulation, respectively), positive (negative) contributions indicate angular momentum that acts to rotate the body toward the ipsilateral (contralateral) leg. Each muscle is depicted using a muscle-specific color to enable comparisons across figures. Unless otherwise indicated, muscles are from the leg specified in the plot title (see Table 1 for muscle group abbreviations). In the nonamputee plots, "contra" indicates the contribution came from the contralateral nonamputee leg.

three legs. As a result, during the first half of stance, the primary contributors to backward angular momentum in the residual, intact, and nonamputee legs were largely similar with GMAX, HAM, and gravity contributing, although HAM was a larger contributor in the residual leg. In addition, during the second half of stance, HAM contributed to backward angular momentum in all three legs. However, VAS contributions throughout stance in the residual leg were opposite that of the nonamputee leg. The importance of the plantarflexors (SOL, GAS) or prosthesis, knee extensors (VAS), and hip extensors (GMAX) to sagittal-plane angular momentum is consistent with their critical role in providing sagittal-plane anteroposterior and vertical propulsion during nonamputee [35,41] and amputee [34] stair ascent while the importance of HAM and RF is consistent with their role in providing anteroposterior propulsion during nonamputee [35] and amputee [34] stair ascent.

Despite the decreased range of sagittal-plane angular momentum in nonamputee stair ascent compared to level walking [3], in both tasks the muscles that extend the ankle, knee, and hip were the most critical to the regulation of sagittal-plane angular momentum [32]. This is consistent with a previous study showing leg extension strength to be the best predictor of an older adult's ability to recover from a trip in the sagittal plane [42]. Similar to both amputee and nonamputee stair ascent, in level walking GMAX, HAM, and gravity generated backward angular momentum while SOL generated forward angular momentum during the first half of stance [32]. However, unlike level walking where GAS generated forward angular momentum during the first half of stance and backward angular momentum during the second half of stance [32], in stair ascent GAS generated forward angular momentum throughout stance. As a result, during the second half of intact and nonamputee stair ascent, HAM became the primary contributor to backward angular momentum in place of GAS while both VAS and HAM contributed to backward angular momentum during the second half of residual stance. Since both GAS and the prosthesis contribute to forward angular momentum throughout stance, these results contradict a previous hypothesis that the increased range of sagittal-plane angular momentum observed during residual stance compared to nonamputee stance could be mitigated if the prosthesis was able to replicate the

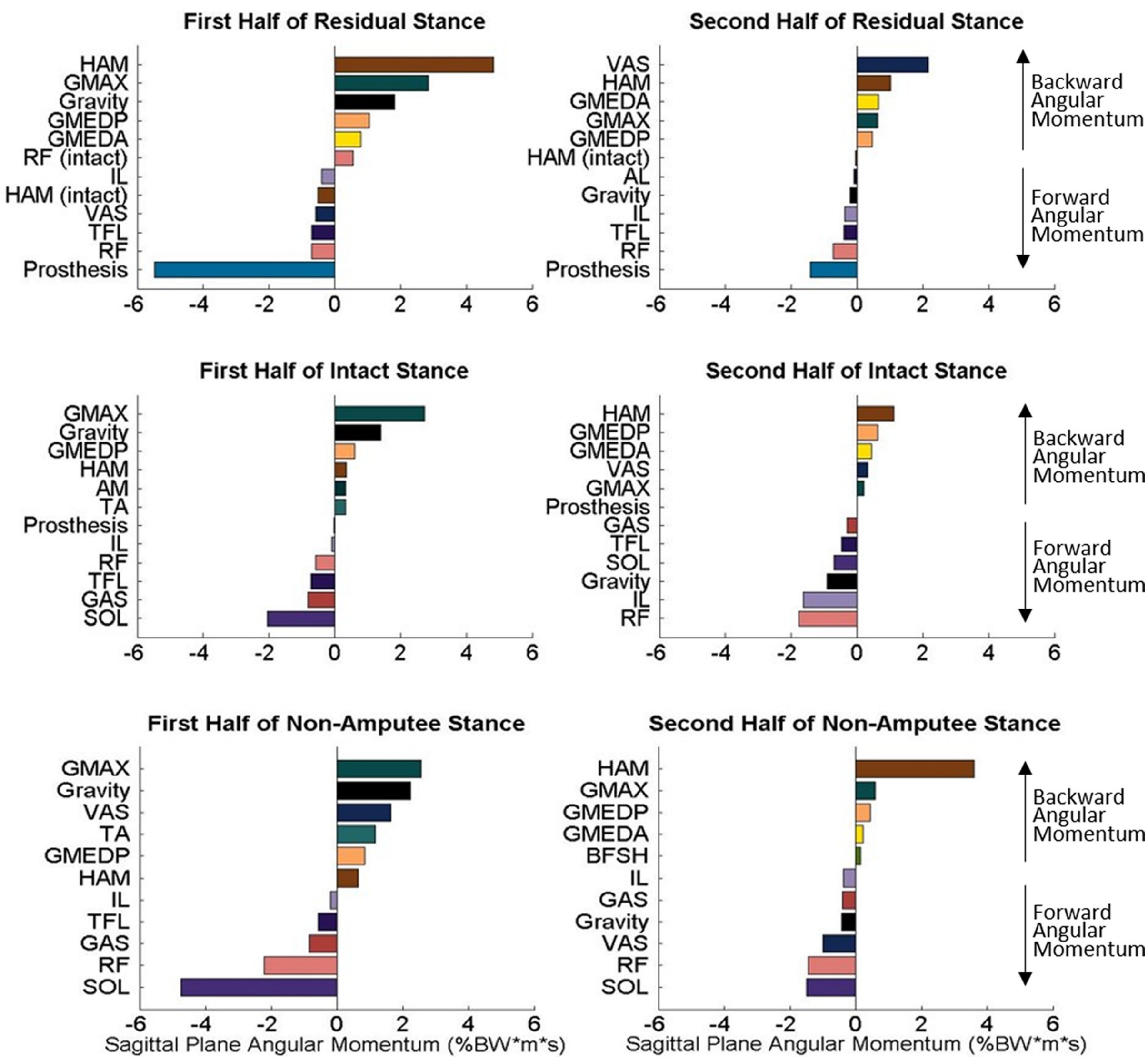

Fig. 3 Primary positive and negative muscle contributions in addition to the contributions from the prosthesis and gravity to whole-body angular momentum in the sagittal plane during the first and second halves of residual, intact, and nonamputee stance during stair ascent. Positive (negative) contributions indicate angular momentum that acts to rotate the body backward (forward). Each muscle is depicted using a muscle-specific color to enable comparisons across figures. Unless otherwise indicated, muscles are from the leg specified in the plot title (see Table 1 for muscle group abbreviations).

function of gastrocnemius [2]. Instead, this emphasizes the need for training programs that focus on improving force production magnitude and timing of the hip and knee extensors (GMAX, HAM, and VAS) which may improve an amputee's ability to control dynamic balance by generating the appropriate backward angular momentum.

In addition, previous studies examining recovery from a sagittal-plane trip found that the increased forward angular momentum generated by tripping over an obstacle during level walking could be restrained by the plantarflexors, hamstrings, and gluteus maximus in the support limb [16] in addition to hip and knee extensor moments in the recovery limb [43]. While this emphasizes the importance of GMAX, HAM, and VAS in generating backward angular momentum, in stair ascent the plantarflexors contribute solely to forward angular momentum. Therefore, the plantarflexors would be unable to actively restrain the increased forward angular momentum following a trip (c.f., during level walking Refs. [16] and [32]) and could only decrease their contributions to forward angular momentum. However, during amputee stair ascent, a passive prosthesis cannot decrease its contribution to forward angular momentum in response to a trip, which is consistent with an amputee's limited ability to respond to balance perturbations in the prosthetic leg [44]. This is further complicated by the increased range of sagittal-plane angular momentum during residual stance compared to nonamputee stance, which suggests decreased dynamic balance control [2]. As a result, if an amputee were to trip during stair ascent it may be more difficult to restrain the sudden increase in forward angular momentum due to the inability of the prosthesis to limit the forward momentum. This further emphasizes the need for training programs to improve force production magnitude and timing of muscles in the intact and residual legs that are capable of providing backward angular momentum (e.g., hip and knee extensors) in amputees.

While the prosthesis replicated the role of the nonamputee plantarflexors in the sagittal plane, it was unable to do so in the frontal plane. In the frontal plane, during the first half of stance the nonamputee plantarflexors contributed to angular momentum that acted to rotate the body toward the ipsilateral leg (ipsilateral angular momentum) while the prosthesis and intact plantarflexors (SOL, GAS) were found to contribute to angular momentum that acted to rotate the body toward the contralateral leg (contralateral angular momentum). To compensate, the hip abductors (GMEDA, GMEDP) increased their contributions to ipsilateral angular momentum during the first half of residual and intact stance compared to nonamputee stance and intact VAS decreased its contribution to contralateral angular momentum. During the second half of stance, the plantarflexors (SOL, GAS) contributed to contralateral angular momentum in both the intact and nonamputee legs, but to a greater extent in the intact leg. To compensate, HAM decreased its contribution to contralateral angular momentum while the hip abductors (GMEDP, GMEDA) increased their contribution to ipsilateral angular momentum in the intact leg. The prosthesis contributed minimally to contralateral angular momentum during the second half of stance in both the residual and intact legs. To compensate for the decreased contribution to contralateral angular momentum provided by the prosthesis, the contribution of gravity to contralateral angular momentum in the residual leg was greater. Despite these altered relative contributions in the residual and intact legs, the range of frontal-plane angular momentum is similar in nonamputee and amputee stair ascent (e.g., Ref. [2]).

In stair ascent, the importance of the plantarflexors (SOL, GAS), prosthesis, hip abductors (GMEDA, GMEDP), HAM, and VAS in the regulation of frontal-plane angular momentum is consistent with their various roles in the frontal-plane tasks of providing vertical propulsion and mediolateral control in amputee [34] and nonamputee [35] stair ascent. Similar to stair ascent, in nonamputee level walking VAS was a primary contributor to contralateral angular momentum during the first half of stance while gluteus medius was the primary contributor to ipsilateral angular momentum throughout stance [33]. In addition, in nonamputee level walking the plantarflexors contributed to contralateral angular momentum [33], similar to the role of the prosthesis and the intact plantarflexors in stair ascent. However, the nonamputee plantarflexors contributed oppositely during the first half of stance. The importance of the hip abductors in generating frontal-plane angular momentum during stair ascent is consistent with the role of the hip abductors in maintaining mediolateral balance during level walking [45–47], in addition to the role of the hip abductor moment in resisting ML perturbations to standing balance [48] and maintaining lateral stability in the frontal plane during stair ascent [49]. It is also consistent with the hip strategy observed in response to mediolateral foot perturbations in both amputees and nonamputees [50].

In general, contributions to angular momentum in each plane were larger during the first half of stance compared to the second half. This suggests during stair ascent, there is an increased demand on the musculature to control angular momentum during the first half of stance in both amputees and nonamputees. Targeted training that improves the force production magnitude and timing of muscles that contribute to angular momentum during the first half of stance may help improve dynamic balance and decrease the risk of falls in both amputees and nonamputees. Emphasis should be placed on the muscles that contribute to backward angular momentum in the sagittal plane where a trip is most likely to occur and the muscles that regulate frontal-plane angular momentum where dynamic balance is more difficult to maintain compared to level walking [2].

One limitation of this study is that validation of musculoskeletal models and simulations is challenging. However, in this study, biomechanically consistent simulations were produced by requiring the simulations to closely emulate experimental data while minimizing unnecessary muscle cocontraction. In addition, simulated muscle excitation timings were previously compared to those available in the literature [34,35]. Because muscle function is state dependent [51], by requiring the state and muscle excitation timings to closely emulate experimental data, we can confidently assess the functional role of the muscles using forward dynamics simulations. Another potential limitation is that the arms were not included in the musculoskeletal model. However, previous studies in level walking demonstrated that the arms contribute negligibly to sagittal- and frontal-plane angular momentum [10] and COM accelerations [45].

Finally, one additional limitation is that group-averaged experimental data were simulated for both amputee and nonamputee stair ascent. However, individuals with amputation have been shown to demonstrate different individual compensations which may not be apparent in the averaged data [52,53]. As a result, future work should focus on generating subject-specific simulations of stair ascent to assess individual deficits in balance control and enable the development of targeted rehabilitation programs tailored to an individual. However, this study is an important first step toward understanding individual muscle contributions to the control of frontal-, transverse- (see Fig. S10 and text available in the Supplemental Materials on the ASME Digital Collection), and sagittal-plane angular momentum in nonamputee and unilateral transtibial amputee stair ascent. These results have important clinical applications in the treatment of movement disorders which exhibit decreased dynamic balance control.

5 Conclusions

This study provides insight into the mechanisms by which individuals with and without amputation control dynamic balance during stair ascent with emphasis on the necessary compensations in individuals with unilateral transtibial amputations. This work has important implications for designing targeted training programs and refined prostheses to improve dynamic balance during stair ascent. The passive prosthesis replicated the role of the nonamputee plantarflexors in the sagittal plane by providing forward angular momentum throughout stance. The prosthesis also largely replicated the role of the nonamputee plantarflexors in the transverse plane but caused a greater change in angular momentum, necessitating additional compensations (see Fig. S10 and text available in the Supplemental Materials on the ASME Digital Collection). The biggest difference between amputee and nonamputee regulation of angular momentum during stair ascent was in the frontal plane, where the prosthesis and nonamputee plantarflexors contributed oppositely during the first half of stance and the prosthesis contributed to a much smaller extent during the second half of stance. This resulted in altered contributions from the intact plantarflexors (SOL, GAS), intact VAS (first half of stance), intact HAM (second half of stance), and the intact and residual hip abductors (GMEDA, GMEDP). Gravity's contribution also increased during residual stance. Therefore, prostheses with reduced contributions to transverse- and frontal-plane angular momentum during the first half of stance could improve balance control during amputee stair ascent and minimize muscle compensations. This could potentially be accomplished by minimizing the prosthesis' contribution to the mediolateral GRF, possibly through frontal-plane stiffness modulation. In addition, targeted training techniques could be implemented to improve force production magnitude and timing of the muscles that primarily contribute to frontal-plane angular momentum where dynamic balance is more difficult to maintain compared to level walking [2]. Additional emphasis could also be placed on training the muscles that contribute to backward angular momentum, which is important when recovering from a trip. Improving force production magnitude and timing of these muscles could help improve dynamic balance control and reduce the risk of a fall during stair ascent in both individuals with and without amputation.

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