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Functional resistance training during walking: Mode of application differentially affects gait biomechanics and muscle activation patterns

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ABSTRACT

Background: Task-specific loading of the limbs—termed as functional resistance training—is commonly used in gait rehabilitation; however, the biomechanical and neuromuscular effects of various forms of functional resistance training have not been studied systematically. This information is crucial for correctly selecting the appropriate mode of functional resistance training when treating individuals with gait disorders.
Research question: To comprehensively evaluate the biomechanical (i.e., joint moment and power) and muscle activation changes with different forms of functional resistance training that are commonly used in clinics and research using biomechanical simulation-based analyses.
Methods: We developed simulations of functional resistance training during walking using OpenSim (Gait2354, 23 degrees of freedom and 54 muscles) and custom MATLAB scripts. We investigated five modes of functional resistance training that have been commonly used in clinics or in research: (1) a weight attached at the ankle, (2) an elastic band attached at the ankle, (3) a viscous device attached to the hip and knee, (4) a weight attached at the pelvis, and (5) a constant backwards pulling force at the pelvis. Lower-extremity joint moments and powers were computed using inverse dynamics and muscle activations were estimated using computed muscle control while walking with each device under multiple resistance levels: normal walking with no resistance, and walking with 30, 60, and 90 Newtons of resistance.
Results: The results indicate that the way in which resistance is applied during gait training differentially affects the internal joint moments, powers, and muscle activations as well as the joints and phase of the gait cycle where the resistance was experienced.
Significance: The results highlight the importance of understanding the joints and muscles that are targeted by various modes of functional resistance training and carefully choosing the best mode of training that meets the specific therapeutic needs of the patient.

1. Introduction

Injuries to the neuro-musculoskeletal system (e.g., stroke, spinal cord injury, lower-extremity injuries, osteoarthritis, etc.) can result in profound gait deficits that can lead to reduced function and long-term health-related quality of life [1–3]. Leg muscle weakness has been consistently linked to abnormal gait patterns and poor biomechanical symmetry in these individuals [4–7]. Accordingly, rehabilitation efforts are often concentrated on restoring muscle strength and control. However, clinical interventions addressing muscle weakness are often performed in a non-functional manner (i.e., exercises performed in a seated or standing position), which may not be optimal for transfer of benefits of training to functional activities, such as walking, because of the phenomenon of "practice specificity" [8–10]. Task-specific loading of the limbs—commonly known as functional resistance training or functional strength training—has been purported as a potential approach to address this issue. Functional resistance training can be applied by having subjects perform weight-bearing exercises (e.g., sit-to-stand, step-ups, etc.) [11–13], as well as by applying additional external loads to the limbs during walking [14–17]. The key advantages of the latter approach are that it is more specific to walking and allows the resistance to be scaled according to the needs of the patient. In the past

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decade, a number of studies have evaluated the short-term and long-term effects of applying external resistive loads during walking on physiological, biomechanical, and clinical outcomes [14–20]. Collectively, these studies have shown that functional resistance training during walking: (1) increases metabolic cost, (2) positively affects biomechanical outcomes, such as muscle activation, joint kinetics, and kinematics, and (3) improves gait function.

It is to be noted, though, that functional resistance training during walking can be performed in multiple ways. For example, studies have used simple devices, such as ankle weights and weighted-vests, to more sophisticated active and passive robotic devices [15–17,20,21]. More importantly, these devices can provide resistance directly to the joints or as end-point forces (e.g., weights tethered to the ankle) and during different phases of gait. Thus, the resulting human-device interactions and muscle activation patterns may differ considerably between these various modes of functional resistance training. Given that functional resistance training is becoming increasingly popular for gait rehabilitation, it is imperative to understand the differential effects of these devices on lower-extremity biomechanics and muscle activation patterns during gait. However, studying the biomechanical effects of these devices in human subjects is challenging because of experimental constraints such as matching resistive loads between devices and subjects, controlling for the confounding effects of fatigue during testing, and studying muscle activation from a large number of lower-extremity muscles.

Computer simulations that utilize musculoskeletal models can be a valuable tool for evaluating the biomechanical effects of different resistive exercises [22]. Previous researchers have used musculoskeletal modeling to compute muscle/joint forces and also to gain insight into how various musculoskeletal properties (e.g., muscle strength, limb lengths, muscle moment arms, etc.) affect biomechanical and functional outcomes (e.g. the kinematics, kinetics, muscle activation, joint loads, jump height, etc.) [22–28]. Such studies have helped in determining movement control during exercise and locomotion [25,28], appropriate kinematics for training [23,24,26,27], and how different exercise schemes affect muscle activation [23,25]. Although previous studies have used musculoskeletal modeling to answer biomechanical questions related to exercise and human locomotion, there are no studies that have comprehensively evaluated how different modes of applying functional resistance training affect gait biomechanics and muscle activation patterns, which may have meaningful implications for how therapy is prescribed in the clinic. Hence, in this study, we used a biomechanical simulation to evaluate the effects of different modes of applying functional resistance training during walking on gait biomechanics and lower-extremity muscle activation patterns during gait.

2. Methods

For this study, we investigated five functional resistance training paradigms (i.e., modes) that have been commonly used in clinics or research: (1) a weight attached at the ankle [20,29], (2) an elastic band attached at the ankle [17], (3) a viscous device attached to the hip and knee (i.e., a device that applies velocity-dependent resistance to the hip and knee joints) [16,30], (4) a weight attached at the pelvis (e.g. weighted vest) [21,31], and (5) a constant backwards pulling force at the pelvis (e.g. pulling a weighted sled) [32,33]. To determine the effects of these various modes of functional resistance training, we devised a simulation-based analysis (Fig. 1) to extract joint moments, powers, and muscle activations while walking with each device under multiple resistance levels: normal walking with no resistance, and walking with 30, 60, and 90 N of resistance. A simulation was ideal for this analysis because it allowed us to maintain exact kinematics across all modes and levels of resistance, monitor activation of muscles that are difficult to measure with surface electromyography, and perform a wide battery of tests without fatiguing a participant.

2.1. Biomechanical simulation in OpenSim

The simulation was run in OpenSim (Version 3.3) using the Gait2354 dynamic musculoskeletal model (23 degrees of freedom and 54 muscles) and modeling tools [34]. We set up the analysis using two marker trajectory files and an external force file provided by OpenSim for this model: the first marker trajectory file tracked a participant (weighing 72.6 kg) standing in a static pose, the second tracked the participant walking over a treadmill for several seconds, and the external force file contained all forces acting on the model during walking (i.e., ground reaction forces). We used the Scale Tool with the static marker file to scale the model to the participant's anthropometry. We then used the Inverse Kinematics Tool with the walking marker file to determine the participant's kinematics during walking. The external force file was then modified based on the mode and level of the resistance (more details on this procedure are provided below in section 2.2). The kinematics and external force files were then fed into the Inverse Dynamics Tool to find joint moments and calculate power, as well as the Computed Muscle Control (CMC) Tool [35] to simulate lower-extremity muscle activation. The maximum isometric force of each muscle in the model was adjusted uniformly (2× default) in order to prevent muscle activation from saturating during CMC while maintaining each muscle's relative contribution to walking.

2.2. Modeling various modes of functional resistance training

Custom MATLAB (version R2017b) programs were written in order to emulate the various modes of applying resistance and modify the external force file (downloadable at http://neurro-lab.engin.umich.edu/downloads). For each mode of resistance, the programs modified the original external force file to (1) include the resistance that was added to the model and (2) adjust the ground reaction forces to balance new external forces. These programs used the kinematics and the governing physical principles of the resistance modes to generate time-varying force information during walking (Fig. 1). The coefficients for each resistance mode (i.e., mass, stiffness, damping, and force constant) were set to provide either 30, 60, or 90 N of resistance to the model, determined as the average force added to the model over the gait cycle. Thus, the amount of resistance applied during walking was similar across all resistance modes. The methods used to calculate these resistances are detailed in the supplemental section. To balance the external forces that were added to the model, a force equal and opposite to the external force was added to the ground reaction forces of the resisted leg during stance. However, simply adding this balancing force to the ground reaction force resulted in large discrete events at heel strike and toe-off. To smoothen these discrete events, the balancing force was multiplied by a scalar that increased linearly from 0 (at heel strike) to 1 (at flat foot) and decreased linearly back to 0 at toe-off (i.e., trapezoidal windowing function). This force balancing was only necessary for the external forces added by the ankle weight, elastic band, pelvis weight, and constant pelvis force. The resistive forces added by the viscous device were internal to the model, and therefore, did not alter ground reaction forces.

2.3. Variables extracted from the simulation

For our analysis, we evaluated the internal joint moments and joint power of the right hip, knee, and ankle in the sagittal plane. In addition to the temporal representation of these variables, we measured the maximum and minimum moments and powers during the stance and swing phases of gait. We also evaluated the muscle activation of the rectus femoris (RF), vastus intermedius (VI), biceps femoris long head (BFL), biceps femoris short head (BFS), gluteus maximus (GMax), tibialis anterior (TA), medial gastrocnemius (MG), soleus (Sol), and gluteus medius (GMed) muscles. The model included three GMed and three GMax muscles, and we used the average activation of these

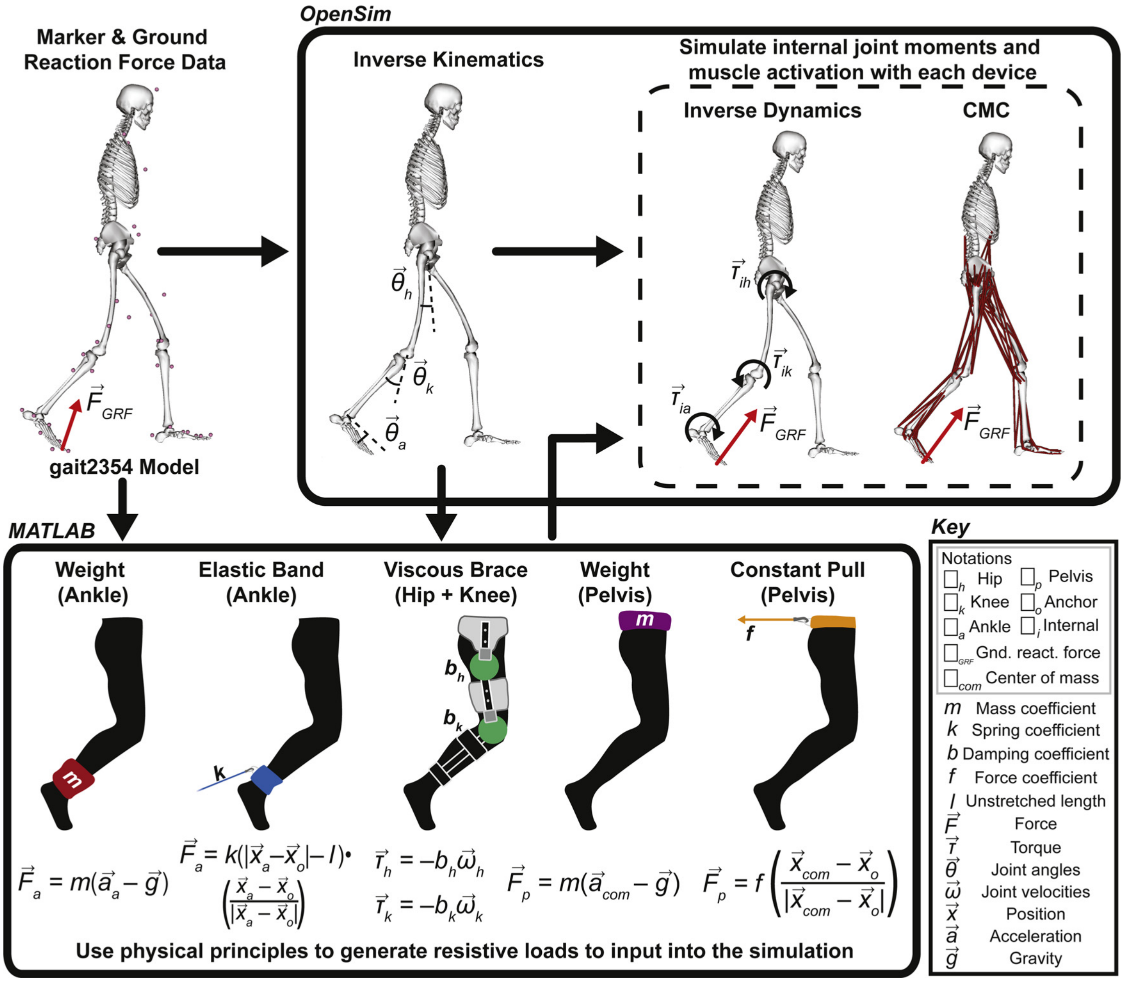

Fig. 1. A schematic of the simulation-based analysis used to estimate the biomechanical effects of walking under various modes of applying resistance. The simulation used the Gait2354 dynamic musculoskeletal model in OpenSim to first generate kinematics from marker trajectories. The kinematics and a file containing the external forces acting on the model were then input into MATLAB to generate time-varying force information for each resistive mode. An updated external force file and the kinematics were then used to run inverse dynamics to generate internal joint moments and power and used in the computed muscle control algorithm (CMC) to generate muscle activations.

muscles in our analyses. For visualization purposes, muscle activation data were depicted as heat maps that highlight the muscles and gait cycle segments where a muscle's activation either increased or decreased when compared with normal walking. In order to highlight only meaningful changes from normal walking, heat maps included muscle activation data only when the product of the percentage change in muscle activation from normal walking and the difference between resisted and normal walking muscle activations exceeded a certain threshold (1.25) (e.g., a 25% increase from baseline with at least a change of 0.05 in activation). The gait cycle was divided into loading response (LR), mid-stance (MSt), terminal stance (TSt), pre-swing (PSw), initial-swing (ISw), mid-swing (MSw), and terminal swing (TSw) to better elucidate when each resistance mode was effective during gait. The raw data for all of the joints and muscles in the model for each mode and level of resistance can be found in the supplemental materials.

3. Results

Traces of the internal joint moments and joint powers, and heatmaps of muscle activations during the 60 N of resistance condition are depicted in Fig. 2. Additionally, the maximum and minimum internal joint moments and powers and the average percent change in muscle activation during the stance and swing phases for all modes and levels of resistance are provided in Tables 1 and 2. The results for the 60 N resistance condition are summarized below. Note that all percentages represent the percentage change of the variable with respect to normal walking with no resistance.

The ankle weight mainly increased hip extension and knee flexion moments (477% and 174%, respectively) and increased power absorption (167%) at the knee joint at the end of the swing phase. The muscle activation of the hamstrings (biceps femoris long and short heads) was increased during the terminal swing (35% and 17%, respectively). This increased hamstring muscle activation corresponded to eccentric contraction of the hamstring muscles because the knee joint was absorbing power over this phase. Interestingly, the ankle weight also reduced the activation of a quadriceps muscle (vastus intermedius) during swing (−48%).

The elastic band increased hip flexion and knee extension moments during the swing phase of gait (852% and 772%, respectively) when compared with normal walking with no resistance. The hip generated more power during the early swing phase (143%) then began absorbing more power towards the end of the swing phase (> 1000%). The knee primarily showed an increase in power generation during the swing phase (> 1000%). The muscle activation of the quadriceps muscles (rectus femoris and vastus intermedius) was increased during the swing phase (198% and 148%, respectively), while that of the hamstring muscles (biceps femoris long and short heads) was decreased during the

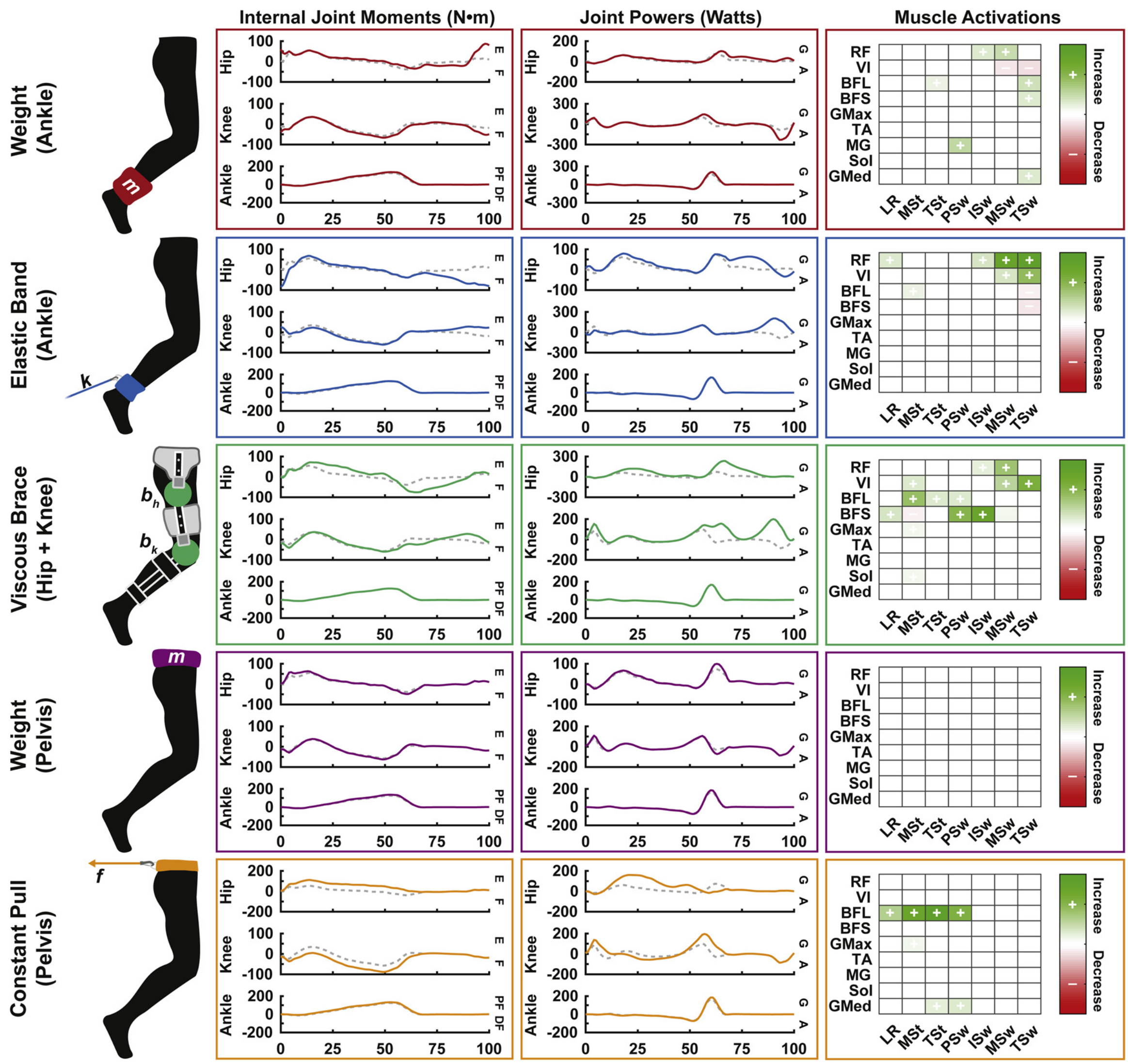

Fig. 2. Simulated joint moments, powers, and muscle activations resulting from common resistance types applied while walking. Plots depict the result of a 60 N resistive force applied using a weight placed at the ankle, an elastic band located at the ankle, a viscous resistance at the hip and knee, a weight placed at the pelvis, and a constant force pulling backwards at the pelvis. Joint moments and powers are plotted against the percentage of the gait cycle, where solid lines represent walking with the load and dashed lines represent normal walking with no resistance. Labels to the right of the moment plots indicate the direction for extension, flexion, plantarflexion, and dorsiflexion. Labels on the power plots indicate a power generation or absorption. Muscle activations are depicted as a heat map of the muscles and the phase of the gait cycle, and indicate a meaningful change (see section 2.3) in muscle activation between resisted and normal walking. Muscle abbreviations: RF (rectus femoris), VI (vastus intermedius), BFL (biceps femoris long head), BFS (biceps femoris short head), GMax (gluteus maximus), TA (tibialis anterior), MG (medial gastrocnemius), Sol (soleus), GMed (gluteus medius). Gait phase abbreviations: LR (loading response), MSt (mid-stance), TSt (terminal stance), PSw (pre-swing), ISw (initial-swing), MSw (mid-swing), TSw (terminal swing).

swing phase (−22% and −9%, respectively).

The viscous device increased hip extension moments during the stance phase (31%) and hip flexion moments during the swing phase (708%) when compared with normal walking with no resistance. In the knee, it increased flexion moments during early swing (18%) and increased extension moments during late swing (759%). The viscous device increased power generation at the hip during the stance and swing phases (219% and 712%, respectively), and also increased power generation at the knee during the swing phase (> 1000%). The muscle activation of the hamstring muscles (biceps femoris long and short heads) was increased during the stance (59% and 32%, respectively) and initial swing phases (20% and 90%, respectively), and that of the quadriceps muscles (rectus femoris and vastus intermedius) was increased during the swing phase (66% and 194%, respectively). These increased quadriceps and hamstring muscle activations corresponded to concentric contractions of these muscles because the hip and knee joints were generating power at these instances.

The weight belt at the pelvis had a minimal effect on gait mechanics at the resistance levels studied in this experiment. The moments, powers, and muscle activation measured while applying resistance closely resembled those of normal walking with no resistance. The most notable change was an increase in knee absorption power during the stance phase (64%). However, quadriceps muscle activation changes during the stance phase were minimal as a result of the added resistance (≤5%).

The backwards force pulling on the pelvis resulted in increased hip extension and knee flexion moments during the stance phase (103% and 54%, respectively) when compared with normal walking with no resistance. Power generation increased at the hip for most of the stance phase (119%) but decreased during the pre-swing phase (−122%). The knee generated more power during the loading response (56%), absorbed more power during mid-stance (112%), then generated more power again during pre-swing (93%). This mode of resistance mostly increased muscle activation of hip extensor muscles (biceps femoris

Table 1
Maximum and minimum sagittal joint moments and powers for each mode of applying functional resistance taining.

Joint Moments (N·m)

		Hip				Knee				Ankle			
		Stance		Swing		Stance		Swing		Stance		Swing	
Mode	R	Max	Min	Max	Min	Max	Min	Max	Min	Max	Min	Max	Min
NW	–	55.1	−38.6	15.1	−8.5	35.2	−57.6	3.2	−19.1	125.4	−11.9	0.8	−2.5
Ankle Weight	30	54.8	−34.3	50.8	−15.1	35.4	−62.2	6.1	−35.7	130.7	−12.8	0.8	−2.3
	60	60.2	−35.0	87.1	−21.8	35.5	−66.7	9.1	−52.4	136.3	−13.9	0.8	−2.2
	90	88.8	−41.2	123.4	−31.9	35.7	−71.2	12.2	−69.0	141.8	−15.0	0.8	−2.2
Elastic Band	30	61.2	−38.9	−11.1	−33.8	29.3	−59.5	11.7	0.7	126.5	−7.5	0.8	−2.4
	60	67.7	−84.7	−15.0	−80.9	26.1	−60.8	27.9	3.8	127.2	−3.2	0.8	−2.4
	90	74.3	−130.6	−18.9	−128.4	46.8	−62.2	46.5	4.6	128.0	−1.1	0.8	−2.3
Viscous Brace	30	63.5	−53.4	19.1	−38.6	35.6	−58.7	12.1	−16.4	125.4	−11.9	0.8	−2.5
	60	72.0	−75.7	23.4	−68.7	36.5	−60.0	27.5	−22.6	125.4	−11.9	0.8	−2.5
	90	84.4	−104.3	27.8	−98.8	37.4	−61.5	43.6	−34.7	125.4	−11.9	0.8	−2.5
Pelvis Weight	30	59.0	−43.8	15.1	−8.9	36.5	−60.1	3.2	−19.1	130.8	−12.1	0.8	−2.4
	60	63.4	−49.1	15.1	−9.2	37.8	−62.6	3.2	−19.1	136.3	−12.3	0.8	−2.3
	90	67.8	−54.3	15.1	−9.6	39.3	−65.1	3.2	−19.1	141.8	−12.5	0.8	−2.3
Pelvis Constant	30	83.3	−17.4	15.1	−7.6	20.0	−73.1	3.2	−19.1	128.9	−9.3	0.8	−2.3
	60	111.6	−6.8	15.1	−6.8	4.7	−88.6	3.2	−19.1	132.5	−7.1	0.8	−2.2
	90	139.8	−3.2	15.1	−6.3	−2.6	−104.1	3.2	−19.1	136.1	−4.8	0.8	−2.3

Joint Powers (Watts)

		Hip				Knee				Ankle			
		Stance		Swing		Stance		Swing		Stance		Swing	
Mode	R	Max	Min	Max	Min	Max	Min	Max	Min	Max	Min	Max	Min
NW	–	73.4	−18.1	26.6	−0.1	100.2	−44.0	5.9	−86.7	163.3	−69.0	2.6	−2.6
Ankle Weight	30	82.6	−18.6	47.4	0.1	119.9	−46.1	11.1	−159.0	182.4	−71.9	2.6	−2.5
	60	102.3	−19.7	68.2	0.3	143.0	−48.1	16.2	−231.9	203.1	−74.8	2.6	−2.3
	90	122.9	−20.8	89.0	0.5	168.8	−50.2	21.4	−305.5	223.8	−77.7	2.6	−2.1
Elastic Band	30	74.7	−16.8	44.0	−13.2	103.7	−31.8	81.3	−15.3	165.0	−69.6	2.6	−2.6
	60	78.9	−16.1	64.7	−34.0	104.9	−35.6	204.7	−19.3	165.1	−70.0	2.6	−2.5
	90	87.7	−15.4	92.7	−54.9	106.1	−70.8	336.9	−23.7	165.2	−70.4	2.6	−2.5
Viscous Brace	30	143.4	−16.0	121.3	0.0	120.6	−36.4	84.6	−26.3	163.3	−69.0	2.6	−2.6
	60	234.2	−14.6	215.9	0.1	156.6	−29.9	199.2	−9.1	163.3	−69.0	2.6	−2.6
	90	329.1	−13.1	310.6	0.2	245.0	−26.7	317.5	−4.7	163.3	−69.0	2.6	−2.6
Pelvis Weight	30	85.3	−20.0	27.8	−0.1	102.3	−51.8	5.9	−86.7	172.7	−72.0	2.6	−2.5
	60	98.5	−22.4	29.1	−0.1	107.8	−72.1	5.9	−86.7	182.1	−75.0	2.6	−2.5
	90	111.8	−24.8	30.3	−0.1	117.7	−92.4	5.9	−86.7	191.5	−78.0	2.6	−2.4
Pelvis Constant	30	107.3	−22.3	24.0	−0.1	143.3	−41.4	5.9	−86.7	172.7	−71.0	2.6	−2.4
	60	160.6	−27.0	21.4	−0.1	193.7	−57.1	5.9	−86.7	182.8	−72.9	2.6	−2.2
	90	220.0	−44.1	18.7	−0.1	246.8	−76.2	5.9	−86.7	193.3	−74.9	2.6	−2.0

Abbreviations: R (resistance level [N]), NW (normal walking with no resistance); Mass of the subject in the model: 72.6 [kg]. For hip and knee joint moments, (+) indicates extension and (−) indicates flexion. For the ankle moments, (+) indicates plantarflexion and (−) indicates dorsiflexion. For all powers, (+) indicates a concentric contraction (power generation) and (−) indicates an eccentric contraction (power absorption).

long head and gluteus medius) during the stance phase (156% and 28%, respectively).

4. Discussion

Functional resistance training is increasing in popularity for gait rehabilitation, and several devices have been used in clinics and research to provide resistance to the leg while walking. However, the resulting human-device interactions and muscle activation patterns may differ considerably based on the mode in which the resistance is applied. An understanding of the biomechanical and neuromuscular effects of different modes of functional resistance training is crucial for tailoring training to patient-specific needs. This is the first study to comprehensively characterize the effects of various modes of functional resistance training on lower-extremity biomechanics and muscle activation during gait using a complex biomechanical simulation-based analysis. We specifically evaluated the effects of an ankle weight, an elastic band attached to the ankle, a viscous device attached to the hip and knee, a weight belt on the pelvis, and a constant backwards pulling force on the pelvis, which are the most commonly used methods in clinics and research. We found that the mode of applying resistance had differential effects on the internal joint moments, powers, and muscle activations—greatly influencing the joints and phase of the gait cycle where the resistance was experienced, as well as the muscles that must counteract the applied resistance.

Our results indicate that, when applying functional resistance training during walking, the mode of resistance can be chosen to account for specific strength deficits or walking impairments. Ankle weights primarily increased internal hip extension and knee flexion moments at the end of the swing phase and provided resistance to the hamstring muscles. An elastic band placed at the ankle increased internal hip flexion and knee extension at the end of the swing phase and provided resistance primarily to the quadriceps muscles. The viscous device at the hip and knee was able to provide resistance to the hamstring muscles over the stance phase and initial swing, and the quadriceps muscles during mid- and late-swing. A weight placed at the pelvis was not very effective at providing resistance during walking. The constant backwards pulling force on the pelvis was the most effective

Table 2
Average percentage change in muscle activation from normal walking for each mode of applying functional resistance training.

Muscle Activation (% Change)											
Mode	R	Phase	RF	VI	BFL	BFS	GMax	TA	MG	Sol	GMed
Ankle Weight	30	Stance	−2	−3	4	0	−6	6	8	−1	−2
		Swing	30	−40	20	1	6	1	4	0	11
	60	Stance	−11	−5	10	2	−6	12	14	2	−2
		Swing	46	−48	35	17	9	2	8	−1	31
	90	Stance	−11	−4	18	7	−3	20	21	8	0
		Swing	69	−53	58	32	2	4	17	1	50
Elastic Band	30	Stance	2	−7	3	0	0	−11	1	2	−1
		Swing	108	75	−17	−2	−11	0	1	0	−12
	60	Stance	47	−10	17	6	9	−18	5	−4	1
		Swing	198	148	−22	−9	−10	0	1	0	−10
	90	Stance	70	−12	22	6	17	−24	8	−6	3
		Swing	281	208	−28	−11	−12	0	1	0	−19
Viscous Brace	30	Stance	−26	15	30	20	11	−5	0	3	3
		Swing	42	114	22	45	6	0	1	−2	−4
	60	Stance	−31	36	59	32	27	−5	0	3	8
		Swing	66	194	20	90	14	1	1	4	−1
	90	Stance	−34	61	92	51	40	−4	1	5	15
		Swing	104	257	27	122	29	3	11	0	5
Pelvis Weight	30	Stance	2	1	−1	−1	1	1	3	4	2
		Swing	9	−2	5	1	0	0	1	1	−2
	60	Stance	5	3	−1	1	1	6	7	13	4
		Swing	9	3	8	4	−1	1	2	2	−1
	90	Stance	7	5	1	1	5	8	11	18	8
		Swing	13	0	12	0	−3	2	3	3	2
Pelvis Constant	30	Stance	−31	−7	67	3	18	−5	6	−5	9
		Swing	4	0	−2	5	−1	0	3	−3	−5
	60	Stance	−32	−4	156	7	38	−14	6	13	28
		Swing	5	0	0	−4	0	2	10	1	5
	90	Stance	−32	−5	249	8	61	−28	5	21	47
		Swing	3	−2	−3	5	3	2	6	2	−3

Abbreviations: R (resistance level [N]), RF (rectus femoris), VI (vastus intermedius), BFL (biceps femoris long head), BFS (biceps femoris short head), GMax (gluteus maximus), TA (tibialis anterior), MG (medial gastrocnemius), Sol (soleus), GMed (gluteus medius). The average percentage change in muscle activation was calculated as: ((resisted − normal)/normal)*100.

mode for resisting hip extension and knee flexion during stance. Overall, these findings were consistent with previous studies on human subjects [14,16,18–20,30,32,33,36,37], albeit, with some exceptions. A key discrepancy was that a constant backwards pulling force on the pelvis has been shown to increase both quadriceps and hamstring activation during stance [32,33], whereas only the hamstring muscles were active in our study. We believe this occurred because participants typically are allowed to adjust their kinematics when walking against a resistance, whereas we constrained the kinematics to normal walking. Thus, the increased hip, knee, and trunk flexion that are commonly observed when walking with a constant backward pulling force could have resulted in more contribution from the quadriceps muscles [32].

An important finding of this study was that a weight placed at the pelvis minimally altered joint moments and muscle activation during resisted walking. Other studies that have investigated this mode of resistance at similar resistance levels have also found that it is ineffective for altering the internal joint moments during walking [20,31]. It is possible that the mass used for these studies was too small to elicit meaningful effects, as studies that have applied much larger resistances (up to 40% of bodyweight) found increases in quadriceps muscle activation [38]. We did not apply resistances this large because we controlled for the amount of external force applied to the model to allow for proper comparison between resistive modes, and we feel masses that large would not be practical for training patients with gait impairments. Similar to a weight placed at the pelvis, a weight placed at the ankle only had a small effect on internal joint moments/powers even with a large mass. Additionally, the increases in internal moments and powers due to a weight placed at the ankle were mainly observed during the terminal swing, which greatly limits how this approach can be applied for functional resistance training during walking.

In our simulation, we constrained the kinematics of the model to match a normal gait pattern without any resistance. Notably, this assumption does not account for kinematic adaptations that would typically occur when resistance is added during walking, which may lead to differences between the simulation results and what would be found through experiments on human subjects. However, there is a strong rationale for constraining the kinematics to normal walking. There is evidence that resisted walking leads to motor slacking [18], where the motor system reduces muscle activation levels and movement excursions to minimize metabolic and movement-related costs [39]. To reduce slacking during functional resistance training, many studies instruct subjects to walk as normally as possible [30,37,40] or provide kinematic feedback [18,19] when the resistance is applied. This helps ensure that the muscles are appropriately loaded so the benefits of functional resistance training can be fully realized. Moreover, studies have shown that, although participants reduce their hip and knee flexion angles during the early part of functional resistance training, they typically adapt to the applied resistance and resume walking with more normal flexion angles with continued training [37]. From these aspects, the model adequately accounted for the effort that the participant will have to exert while walking with resistance and resembled the way in which functional resistance training is performed in the clinic or research.

5. Limitations

There are some potential limitations to this study. We used a generic musculoskeletal model based on healthy participants in our simulations, which limits the generalizability of the findings to clinical populations (e.g., stroke) that often present with altered biomechanics and muscle weakness. However, it is to be noted that clinical populations with gait deficiencies share many kinematic gait features with normal walking (e.g., the leg must swing forward and backward throughout the gait cycle to complete a stride); hence, we believe that many of the general findings presented in this study will carry over to clinical populations. Another limitation is that we do not know how much change in muscle activation or moment data is needed to induce meaningful physiological adaptations (e.g., increase in gait speed). However, the observed biomechanical changes are in general similar to or greater than those reported in the literature [20,32,36,41,42]—a change that has been shown to positively affect gait outcomes in patient populations after a course of an intervention [15,17,29]. There are also several assumptions made during musculoskeletal modeling. For example, biomechanical simulations typically rely on a minimum effort principle (e.g., the sum of the squared muscle activations, as done in this study) to simulate muscle activation. While this approximation of a human's motor control objective may be reasonable for a healthy population, it may not hold when predicting the muscle activation of a pathological gait, which can be influenced by other factors such as pain or fatigue. In these cases, patients may alter their motor control to more heavily favor stability or preserve joint integrity by increasing antagonist muscle activity, which a minimum effort principle fails to predict [22,43]. As a result, caution should be exercised when generalizing the study results to a patient population.

6. Conclusion

In summary, using a simulation-based analysis we show that the mode of applying resistance greatly affects joint moments, powers, and muscle activation, as well as the phase of the gait cycle where the resistance was experienced. Specifically, we show that providing resistance via an elastic band at the ankle can be used to isolate and target the quadriceps muscles during the swing phase, whereas a viscous hip and knee device could be used to target both hip/knee flexors and extensors. A constant backward pulling force at the pelvis could be used to primarily target the hip extensors during the stance phase. A weight placed at the pelvis or at the ankle minimally altered joint moments and muscle activation during resisted walking. Thus, the detailed biomechanical and muscle activation changes described in this study can be used to guide rehabilitation so that resistances for functional resistance training during walking can be prescribed to better account for patient-specific strength deficits or walking impairments.

Declaration of Competing Interest

None.

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Appendix A. Supplementary data

Supplementary material related to this article can be found, in the online version, at doi:https://doi.org/10.1016/j.gaitpost.2019.10.024.

References

[1] A. Moseley, A. Wales, R. Herbert, K. Schurr, S. Moore, Observation and analysis of hemiplegic gait: stance phase, Aust. J. Physiother. 39 (1993) 259–267.

[2] J.E. Naili, P. Wretenberg, V. Lindgren, M.D. Iversen, M. Hedstrom, E.W. Brostrom, Improved knee biomechanics among patients reporting a good outcome in knee-related quality of life one year after total knee arthroplasty, BMC Musculoskelet. Disord. 18 (2017) 122.

[3] B. Pietrosimone, J.T. Blackburn, D.A. Padua, S.J. Pfeiffer, H.C. Davis, B.A. Luc-Harkey, et al., Walking gait asymmetries 6 months following anterior cruciate ligament reconstruction predict 12-month patient-reported outcomes, J. Orthop. Res. 36 (2018) 2932–2940.

[4] J.T. Blackburn, B. Pietrosimone, M.S. Harkey, B.A. Luc, D.N. Pamukoff, Quadriceps function and gait kinetics after anterior cruciate ligament reconstruction, Med. Sci. Sports Exerc. 48 (2016) 1664–1670.

[5] M. Lewek, K. Rudolph, M. Axe, L. Snyder-Mackler, The effect of insufficient quadriceps strength on gait after anterior cruciate ligament reconstruction, Clin Biomech (Bristol, Avon). 17 (2002) 56–63.

[6] B. Lindmark, E. Hamrin, Relation between gait speed, knee muscle torque and motor scores in post-stroke patients, Scand. J. Caring Sci. 9 (1995) 195–202.

[7] R. Nakamura, T. Hosokawa, I. Tsuji, Relationship of muscle strength for knee extension to walking capacity in patients with spastic hemiparesis, Tohoku J. Exp. Med. 145 (1985) 335–340.

[8] G. Williams, M. Kahn, A. Randall, Strength training for walking in neurologic rehabilitation is not task specific: a focused review, Am. J. Phys. Med. Rehabil. 93 (2014) 511–522.

[9] M.C. Morrissey, E.A. Harman, M.J. Johnson, Resistance training modes: specificity and effectiveness, Med. Sci. Sports Exerc. 27 (1995) 648–660.

[10] L. Proteau, R.G. Marteniuk, L. Levesque, A sensorimotor basis for motor learning: evidence indicating specificity of practice, Q. J. Exp. Psychol. A 44 (1992) 557–575.

[11] A. Kerr, A. Clark, E.V. Cooke, P. Rowe, V.M. Pomeroy, Functional strength training and movement performance therapy produce analogous improvement in sit-to-stand early after stroke: early-phase randomised controlled trial, Physiotherapy 103 (2017) 259–265.

[12] H. Lohne-Seiler, M.K. Torstveit, S.A. Anderssen, Traditional versus functional strength training: effects on muscle strength and power in the elderly, J. Aging Phys. Act. 21 (2013) 51–70.

[13] S.W. Blundell, R.B. Shepherd, C.M. Dean, R.D. Adams, B.M. Cahill, Functional strength training in cerebral palsy: a pilot study of a group circuit training class for children aged 4-8 years, Clin. Rehabil. 17 (2003) 48–57.

[14] C. Duclos, S. Nadeau, N. Bourgeois, L. Bouyer, C.L. Richards, Effects of walking with loads above the ankle on gait parameters of persons with hemiparesis after stroke, Clin Biomech (Bristol, Avon). 29 (2014) 265–271.

[15] T. Lam, K. Pauhl, A. Ferguson, R.N. Malik, Krassioukov A. Bkin, et al., Training with robot-applied resistance in people with motor-incomplete spinal cord injury: pilot study, J. Rehabil. Res. Dev. 52 (2015) 113–129.

[16] E.P. Washabaugh, C. Krishnan, A wearable resistive robot facilitates locomotor adaptations during gait, Restor. Neurol. Neurosci. 36 (2018) 215–223.

[17] M. Wu, J. Kim, D.J. Gaebler-Spira, B.D. Schmit, P. Arora, Robotic resistance treadmill training improves locomotor function in children with cerebral palsy: a randomized controlled pilot study, Arch. Phys. Med. Rehabil. 98 (2017) 2126–2133.

[18] E.P. Washabaugh, E.S. Claflin, R.B. Gillespie, C. Krishnan, A novel application of eddy current braking for functional strength training during gait, Ann. Biomed. Eng. 44 (2016) 2760–2773.

[19] T. Klarner, J.S. Blouin, M.G. Carpenter, T. Lam, Contributions to enhanced activity in rectus femoris in response to Lokomat-applied resistance, Exp. Brain Res. 225 (2013) 1–10.

[20] R.C. Browning, J.R. Modica, R. Kram, A. Goswami, The effects of adding mass to the legs on the energetics and biomechanics of walking, Med. Sci. Sports Exerc. 39 (2007) 515–525.

[21] M.L. Puthoff, B.J. Darter, D.H. Nielsen, H.J. Yack, The effect of weighted vest walking on metabolic responses and ground reaction forces, Med. Sci. Sports Exerc. 38 (2006) 746–752.

[22] F. Schellenberg, K. Oberhofer, W.R. Taylor, S. Lorenzetti, Review of modelling techniques for in vivo muscle force estimation in the lower extremities during strength training, Comput. Math. Methods Med. 2015 (2015) 483921.

[23] M. Pluss, F. Schellenberg, W.R. Taylor, S. Lorenzetti, Towards subject-specific strength training design through predictive use of musculoskeletal models, Appl. Bionics Biomech. 2018 (2018) 9721079.

[24] B.J. Fregly, C.D. Fregly, B.T. Kim, Computational prediction of muscle moments during ARED squat exercise on the international space station, J. Biomech. Eng. 137 (2015) 121005.

[25] T.W. Dorn, J.M. Wang, J.L. Hicks, S.L. Delp, Predictive simulation generates human adaptations during loaded and inclined walking, PLoS One 10 (2015) e0121407.

[26] R.F. Escamilla, N. Zheng, T.D. Macleod, W.B. Edwards, R. Imamura, A. Hreljac, et al., Patellofemoral joint force and stress during the wall squat and one-leg squat, Med. Sci. Sports Exerc. 41 (2009) 879–888.

[27] C.L. Lewis, S.A. Sahrmann, D.W. Moran, Effect of position and alteration in synergist muscle force contribution on hip forces when performing hip strengthening exercises, Clin Biomech (Bristol, Avon). 24 (2009) 35–42.

[28] M.G. Pandy, F.E. Zajac, E. Sim, W.S. Levine, An optimal control model for maximum-height human jumping, J. Biomech. 23 (1990) 1185–1198.

[29] T. Lam, K. Luttmann, A. Houldin, C. Chan, Treadmill-based locomotor training with leg weights to enhance functional ambulation in people with chronic stroke: a pilot study, J. Neurol. Phys. Ther. 33 (2009) 129–135.

[30] T. Lam, M. Wirz, L. Lunenburger, V. Dietz, Swing phase resistance enhances flexor muscle activity during treadmill locomotion in incomplete spinal cord injury, Neurorehabil. Neural Repair 22 (2008) 438–446.

[31] A.J. Kubinski, J.S. Higginson, Strategies used during a challenging weighted walking task in healthy adults and individuals with knee osteoarthritis, Gait Posture 35 (2012) 6–10.

[32] K.R. Mun, B.B.S. Yeo, Z. Guo, S.C. Chung, H. Yu, Resistance training using a novel robotic walker for over-ground gait rehabilitation: a preliminary study on healthy subjects, Med. Biol. Eng. Comput. 55 (2017) 1873–1881.
[33] M. Lawrence, E. Hartigan, C. Tu, Lower limb moments differ when towing a weighted sled with different attachment points, Sports Biomech. 12 (2013) 186–194.
[34] S.L. Delp, J.P. Loan, M.G. Hoy, F.E. Zajac, E.L. Topp, J.M. Rosen, An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures, IEEE Trans. Biomed. Eng. 37 (1990) 757–767.
[35] D.G. Thelen, F.C. Anderson, S.L. Delp, Generating dynamic simulations of movement using computed muscle control, J. Biomech. 36 (2003) 321–328.
[36] D.N. Savin, S.C. Tseng, S.M. Morton, Bilateral adaptation during locomotion following a unilaterally applied resistance to swing in nondisabled adults, J. Neurophysiol. 104 (2010) 3600–3611.
[37] T. Lam, M. Anderschitz, V. Dietz, Contribution of feedback and feedforward strategies to locomotor adaptations, J. Neurophysiol. 95 (2006) 766–773.
[38] K.M. Simpson, B.J. Munro, J.R. Steele, Backpack load affects lower limb muscle activity patterns of female hikers during prolonged load carriage, J. Electromyogr. Kinesiol. 21 (2011) 782–788.
[39] D.J. Reinkensmeyer, O. Akoner, D.P. Ferris, K.E. Gordon, Slacking by the human motor system: computational models and implications for robotic orthoses, Conf. Proc. IEEE Eng. Med. Biol. Soc. 2009 (2009) 2129–2132.
[40] D. Barthelemy, S. Alain, M.J. Grey, J.B. Nielsen, L.J. Bouyer, Rapid changes in corticospinal excitability during force field adaptation of human walking, Exp. Brain Res. 217 (2012) 99–115.
[41] A. Houldin, K. Luttin, T. Lam, Locomotor adaptations and aftereffects to resistance during walking in individuals with spinal cord injury, J. Neurophysiol. 106 (2011) 247–258.
[42] S.C. Yen, J.M. Landry, M. Wu, Size of kinematic error affects retention of locomotor adaptation in human spinal cord injury, J. Rehabil. Res. Dev. 50 (2013) 1187–1200.
[43] G. Bergmann, F. Graichen, A. Rohlmann, Hip joint contact forces during stumbling, Langenbecks Arch. Surg. 389 (2004) 53–59.