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Muscle work is increased in pre-swing during hemiparetic walking

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ABSTRACT

Background: Muscle mechanical work is likely affected by gait abnormalities in hemiparetic walking during the paretic pre-swing phase (i.e., double support phase preceding paretic toe-off). Previous experimental studies suggest that muscle work may be decreased in the paretic leg, but paretic work may have been underestimated since experimental approaches based on net joint moments do not account for co-contraction of antagonist muscles. Also, whether the non-paretic leg does more work compared to control subjects at matched speeds and how work generation may differ between hemiparetic subjects walking with different self-selected speeds remains unknown.

Methods: Three-dimensional forward dynamics simulations of two representative hemiparetic subjects walking with different self-selected speeds (i.e., limited community = 0.45 m/s and community walkers = 0.9 m/s) and a speed and age-matched control subject were generated to quantify musculotendon (fiber and in-series tendon) work during paretic pre-swing.

Findings: Total paretic and non-paretic fiber work were increased in both the limited community and community hemiparetic walkers compared to the control. Increased fiber work in the limited community walker was primarily related to decreased fiber and tendon work by the paretic plantar flexors requiring compensatory work by other muscles. Increased fiber work in the community walker was primarily related to increased work by the hip abductors and adductors.

Interpretation: The hemiparetic walkers would expend more metabolic energy during pre-swing if the hemiparetic and control subjects were to perform work with the same mechanical efficiency. These results may partly explain the increased metabolic cost of hemiparetic walkers compared to nondisabled walkers at matched speeds.

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1. Introduction

Muscle mechanical work, a quantity that cannot be directly measured during walking, is likely affected by abnormalities observed in hemiparetic walking. Increased co-contraction between antagonist muscles (Knutsson and Richards, 1979; Perry, 1993) and increased reliance on medial–lateral muscles for stability compared to non-disabled subjects (Corriveau et al., 2004; Niam et al., 1999) each can result in increased mechanical work without a corresponding increase in walking speed. Furthermore, increased stance limb knee flexion observed in the paretic leg of some hemiparetic subjects (De Quervain et al., 1996) likely increases mechanical work of the paretic muscles, as the knee extensor muscles have a lower mechanical advantage in a flexed-knee position and must generate greater forces to support body weight when the stance limb is more flexed during walking (Biewener et al., 2004). Recently, a simulation analysis of nondisabled walking found that musculotendon (i.e., muscle fiber and in-series tendon) work was consistent with metabolic cost during each region of the gait cycle (Umberger, 2010). Since musculotendon work may partly explain the increased metabolic cost of walking in hemiparetic subjects compared to nondisabled controls at matched speeds (Zamparo et al., 1995), understanding differences in musculotendon work between hemiparetic and nondisabled subjects is important.

Experimental studies suggest that muscle work is likely decreased in the paretic leg. Previously, Olney et al. (1991) reported that the paretic leg performed less positive and negative joint moment work compared to the non-paretic leg. Chen and Patten (2008) compared joint work produced by the paretic hip, knee and ankle during paretic leg pre-swing to age and speed-matched control subjects and found that paretic ankle plantar flexor work was greatly reduced compared to the controls. Increased paretic hip flexor and knee extensor moment work partially offset the reduction in paretic ankle work, but the net joint moment work in the paretic leg was still significantly reduced compared to nondisabled controls. However, paretic work is likely underestimated by net joint moment work since this experimental approach does not account for co-contraction of antagonistic

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muscles (Sasaki et al., 2009), which may be increased in the paretic leg. Also, it remains unknown whether the non-paretic leg does more work compared to control subjects at matched speeds and how work generation may differ between hemiparetic subjects walking with different self-selected speeds.

Other studies have investigated mechanical work by computing external work (i.e., work done to move the body center-of-mass) and internal work (work done to move the body segments relative to the body center-of-mass). However, external and internal work are not independent (Kautz and Neptune, 2002) and only indirectly related to the mechanical energetic cost of muscles (Sasaki et al., 2009). In addition, both external and internal work are limited in their ability to account for co-contraction and elastic energy storage and return (Neptune et al., 2004; Sasaki et al., 2009). One approach to overcome these limitations is to use modeling and simulation techniques to estimate individual musculotendon work that can be partitioned into positive and negative work. In addition, musculotendon work can be partitioned into fiber and tendon work, which is important because tendons presumably do work while incurring little to no metabolic cost, whereas positive and negative work done by muscle fibers constitutes the majority of the metabolic energy expended by muscles.

The purpose of this study was to use musculoskeletal modeling and simulation analyses to compare individual musculotendon work generated by representative hemiparetic subjects classified according to functional walking status (Perry et al., 1995) (i.e., limited community = 0.4–0.8 m/s and community walkers = >0.8 m/s) to a representative speed and age-matched control subject. Musculotendon work performed by representative hemiparetic walkers at two different speeds was analyzed because metabolic cost is more increased compared to speed-matched controls for hemiparetic walkers with slower self-selected speeds than for hemiparetic walkers with faster self-selected speeds (Zamparo et al., 1995). As a first step in understanding differences between the hemiparetic and control subjects, the paretic leg pre-swing phase was analyzed because many abnormalities secondary to stroke that limit walking speed occur during this important double support phase of the gait cycle (Chen et al., 2005; De Quervain et al., 1996).

2. Methods

2.1. Experimental data

Experimental data were collected from 51 hemiparetic subjects walking at self-selected speeds without use of an assistive device or ankle–foot orthosis and 21 nondisabled elderly subjects walking at self-selected speeds and speeds of 0.3, 0.6 and 0.9 m/s as part of a larger study (Peterson et al., 2010b). A safety harness mounted to the ceiling that provided no body weight support protected the subjects in the event of loss of balance. All subjects signed informed consent and the Institutional Review Boards of the University of Florida and the University of Texas approved the protocol. Three-dimensional (3D) ground reaction forces (GRFs) and kinematics, and bilateral EMG from eight lower limb muscles (medial gastrocnemius, soleus, tibialis anterior, rectus femoris, vastus lateralis, biceps femoris, semimembranosus, and gluteus medius) were recorded at 2000 Hz, 100 Hz and 2000 Hz, respectively, during 30 s walking trials on a split-belt instrumented treadmill (Tecmachine). Raw kinematic and GRF data were low-pass filtered with a fourth-order zero-lag Butterworth filter with cutoff frequencies of 6 Hz and 20 Hz, respectively, using Visual 3D (C-Motion, Inc.). All data were time normalized to 100% of the paretic (ipsilateral, right for control) gait cycle and averaged across consecutive gait cycles within each subject at each speed.

From this data set, we selected a representative subject from the limited community walker and community walker functional groups by choosing the person with the average percent of paretic propulsion value closest to the functional group average (Bowden et al., 2006). These subjects (limited community: male, left hemiparesis, single cerebrovascular infarction, age = 53 years, time post stroke = 2 years 1 month, self-selected treadmill speed = 0.45 m/s; community: male, left hemiparesis, single cerebrovascular infarction, age = 60 years, time post stroke = 8 years 5 months, self-selected treadmill speed = 0.9 m/s) and an age matched control subject (female, age = 59 years) walking at speeds of 0.6 and 1.0 m/s were selected for the simulation analyses. From the subject's individual trials, the individual gait cycle with the minimum root-mean-square difference in joint angles and GRFs compared to the subject's average data was used as tracking data in the optimization cost function (see Dynamic optimization below).

2.2. Musculoskeletal model

The previously described 3D musculoskeletal model (Peterson et al., 2010a) had 23 degrees-of-freedom and was developed using SIMM (MusculoGraphics, Inc.) with musculoskeletal geometry based on Delp et al. (1990). The model was driven by 43 Hill-type musculotendon actuators per leg with muscle contraction dynamics governed by Hill-type properties (Zajac, 1989) and activation dynamics modeled using a nonlinear first-order differential equation (Raasch et al., 1997). Muscle specific activation and deactivation time constants were derived from Winters and Stark (1988). Polynomial equations were used to estimate musculotendon lengths and moment arms (Menegaldo et al., 2004). The foot-ground contact was modeled with 31 independent visco-elastic elements on the bottom of each foot (Neptune et al., 2000). Passive torques representing the forces applied by ligaments, passive tissue and joint structures were applied at each joint (Anderson, 1999; Audu and Davy, 1985). The dynamical equations-of-motion were derived using SD/FAST (PTC).

2.3. Dynamic optimization

Forward dynamics simulations from paretic mid-stance to paretic toe-off (right leg for control simulations) were generated using a simulated annealing optimization algorithm (Goffe et al., 1994) that varied the muscle excitation patterns until the cost function was minimized. The cost function (Eq. (1)) computed differences between simulated and experimentally measured walking data (23 degrees-of-freedom and 3 GRFs) normalized by the standard deviation (SD) of the experimental walking data at each time step i. Total muscle stress (muscle force (F) divided by muscle cross-sectional area (A) summed across 86 muscles) was also included in the cost function to minimize co-contraction while reproducing the experimental kinematics and GRFs equally well (Eq. (1)). Bimodal patterns were used to define the muscle excitations that were described by the onset, offset and amplitude of each mode (Peterson et al., 2010a).

$$J = \sum_{i=1}^{t}\left[\sum_{j=1}^{26}\frac{\left(Exp_{ij}-Sim_{ij}\right)^2}{SD_{ij}^2} + \sum_{m=1}^{86}\frac{F_{im}}{A_{im}}\right] \qquad (1)$$

2.4. Work calculations

Individual musculotendon (MT), fiber (parallel active and passive components) and tendon (series-elastic element) power were computed as the product of corresponding force and velocity vectors obtained from the Hill-type muscle model. Net MT, fiber and tendon work were computed by integrating the corresponding power over the paretic (ipsilateral for control) pre-swing phase. Positive and negative work were computed by integrating the positive and negative portions of the power trajectories over pre-swing and summed (i.e., positive work plus absolute value of negative work) to determine the total MT, fiber and tendon work. Work quantities were summed for muscles that were grouped according to anatomical classification (see Table 1).

3. Results

The simulated GRFs and kinematics of the limited community and community hemiparetic walkers and speed-matched controls from mid-stance to toe-off emulated well the experimental data (Peterson et al., 2010a). Joint angles and normalized ground reaction forces were nearly always within "two standard deviations" of the experimental data with an average error of 1.77 degrees and 2.89 %BW, respectively. The right and left legs of the control simulations are referred to as the ipsilateral and contralateral legs, respectively, for comparison with the paretic and non-paretic legs. Paretic (ipsilateral for control) and non-paretic (contralateral for control) muscle excitation timings compared well with the experimental EMG (Figs. 1 and 2) and with data available in the literature (Den Otter et al., 2004, 2007).

3.1. Musculotendon work by the limited community hemiparetic walker and control

The limited community hemiparetic walker had increased total (absolute sum of positive and negative) paretic leg MT work during pre-swing compared to the ipsilateral leg of the control subject walking at 0.6 m/s (Fig. 3A). Negative paretic MT work was increased and positive paretic MT work was decreased, such that the net paretic MT work was decreased compared to the ipsilateral leg (Fig. 3A). Primary differences between paretic and ipsilateral leg fiber work occurred in the vasti muscle group (VAS), rectus femoris (RF), iliopsoas (IL), sartorius (SAR) and the plantar flexors (Fig. 4A). Paretic VAS and RF had increased negative fiber work in the limited community walker relative to the ipsilateral control leg (Fig. 4A). Also, paretic IL and SAR had increased positive fiber work compared to the ipsilateral control leg (Fig. 4A). Positive tendon work was decreased in the paretic leg of the limited community walker compared to the ipsilateral control leg (Fig. 3A) primarily due to decreased positive tendon work by paretic gastrocnemius (GAS), but also by paretic soleus and other plantar flexors (SOL) (Fig. 4A).

Net and total non-paretic leg MT work during paretic pre-swing were increased relative to the contralateral control leg (Fig. 3B). Total fiber and tendon work were increased in the non-paretic leg largely because non-paretic positive fiber work and negative tendon work were increased relative to the contralateral control leg (Fig. 3B). Non-paretic and contralateral gluteus maximus (GMAX) and hamstrings (HAM) did much positive work in the limited community walker and control, respectively (Fig. 4B). Non-paretic SOL and GAS had increased positive fiber work and negative tendon work relative to the contralateral control leg (Fig. 4B).

Table 1
The 43 musculotendon actuators per leg were combined into 18 groups according to their anatomical classification.

Muscle name	Group
Iliacus, psoas	IL
Adductor longus, adductor brevis, pectineus	AL
Quadratus femoris	QFEM
Superior, middle and inferior adductor magnus	AM
Sartorius	SAR
Rectus femoris	RF
Vastus medialis, lateralis, and intermedius	VAS
Anterior, middle and posterior gluteus medius	GMED
Piriformis	PIRI
Gemellus	GEM
Anterior, middle and posterior gluteus minimus	GMIN
Tensor fascia lata	TFL
Anterior, middle, and posterior gluteus maximus	GMAX
Semitendinosus, semimembranosus, gracilis, biceps femoris long head	HAM
Biceps femoris short head	BFSH
Medial and lateral gastrocnemius	GAS
Soleus, tibialis posterior, peroneus brevis, flexor digitorum longus, flexor hallucislongus	SOL
Tibialis anterior, extensor digitorum longus, peroneus tertius, extensor hallucislongus	TA

3.2. Musculotendon work by the community hemiparetic walker and control

Paretic leg net MT work by the community hemiparetic walker during pre-swing was similar to the ipsilateral leg of the control subject walking at 1.0 m/s, though the total paretic MT work was decreased (Fig. 5A). The paretic leg had increased positive fiber work compared to the ipsilateral control leg (Fig. 5A), primarily due to increased positive fiber work by paretic SOL and GAS, and also by gluteus medius (GMED), gluteus minimus (GMIN) and adductor magnus (AM) (Fig. 6A). Negative fiber work by paretic quadratus femoris (QFEM) was increased compared to the ipsilateral control leg (Fig. 6A). Positive tendon work by paretic GAS was much decreased relative to the ipsilateral GAS (Fig. 6A). Paretic leg negative tendon work in the community walker was increased relative to the ipsilateral control leg (Fig. 5A) since paretic AM, GMIN and GMED generated negative tendon power during much of the paretic pre-swing phase (Fig. 6A).

Net and total non-paretic leg MT work by the community hemiparetic walker during paretic pre-swing were decreased compared to the contralateral control leg (Fig. 5B). Net fiber and tendon work were similar between the non-paretic and contralateral legs, although the total non-paretic fiber and tendon work were increased for the community walker relative to the contralateral control leg (Fig. 5B). Non-paretic GMED, GMIN and piriformis (PIRI) did more fiber work and non-paretic VAS, GMAX and HAM did less fiber work during pre-swing compared to the contralateral control leg (Fig. 6B).

4. Discussion

The purpose of this study was to compare individual musculotendon (MT) work generated by limited community and community hemiparetic walkers to speed and age-matched controls during pre-swing. For the limited community walker, net MT work by the paretic leg was decreased relative to the ipsilateral leg of the control subject walking at 0.6 m/s, which was consistent with an experimental study that found decreased net paretic joint moment work during paretic pre-swing by hemiparetic subjects compared to speed-matched controls (Chen and Patten, 2008). However, total paretic and non-paretic fiber work was increased. For the community walker whose self-selected speed was similar to that of the age-matched control, net paretic MT work was similar to the ipsilateral leg of the control subject. However, similar to the limited community walker, total paretic and non-paretic fiber work were increased. Thus, if the hemiparetic walkers and control subjects perform work with the same mechanical efficiency as suggested by previous studies (Detrembleur et al., 2003; Stoquart et al., 2005), the hemiparetic walkers would expend more metabolic energy during paretic pre-swing compared to the controls during ipsilateral pre-swing.

4.1. Limited community walker: Increased work related to paretic plantar flexor deficits

More fiber work was required from paretic and non-paretic muscles of the limited community walker to achieve a similar speed as the control largely due to paretic plantar flexor deficits. Consistent with experimental and simulation studies of nondisabled walking (Ishikawa et al., 2005; Neptune et al., 2008), SOL and GAS recovered

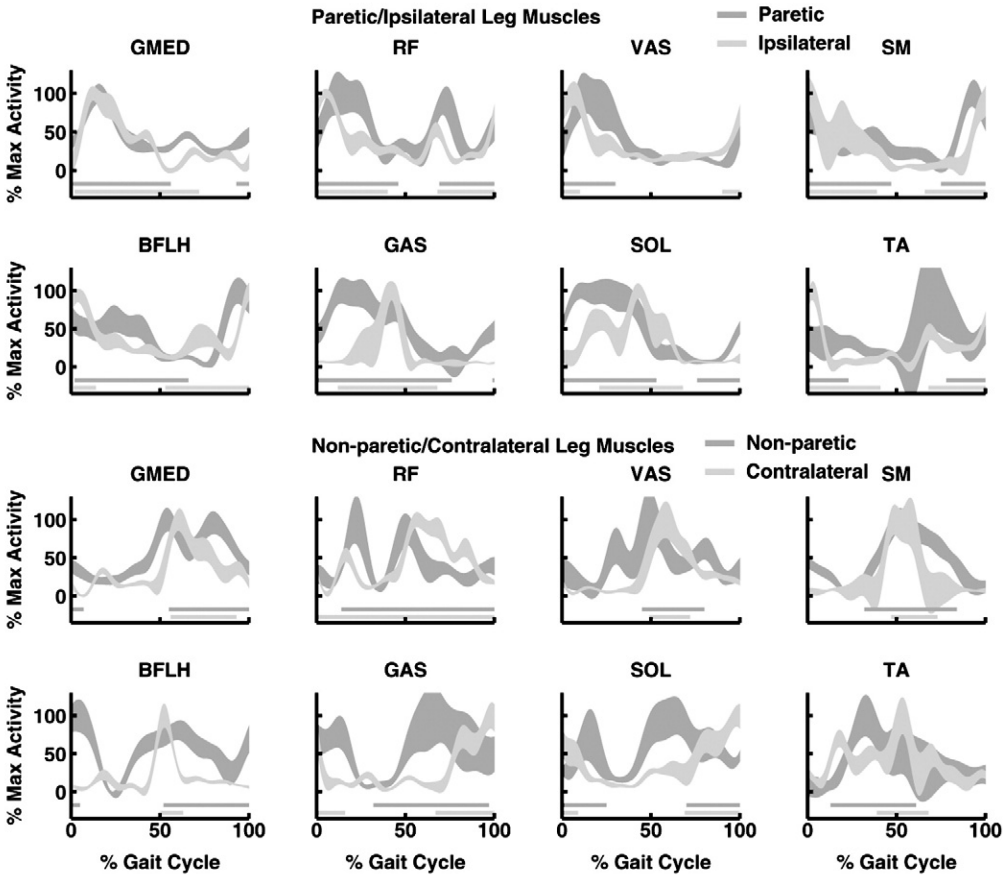

Fig. 1. For the limited community walker and control subject at 0.6 m/s, paretic (ipsilateral for control) and non-paretic (contralateral for control) EMG and optimized excitation timings (bars below the EMG indicate when a muscle is excited) normalized to the paretic leg (ipsilateral for control) gait cycle. Average EMG is shown with ± one standard deviation of the 30 s walking trial. Muscle excitations were optimized from mid-stance to toe-off corresponding to 36 to 84% of the paretic gait cycle for the limited community walker and 33 to 72% of the ipsilateral gait cycle for the control subject. The abbreviated muscle names are: GMED, gluteus medius; RF, rectus femoris; VAS, vastus lateralis; SM, semimembranosus; BFLH, biceps femoris long head; GAS, medial gastrocnemius; SOL, soleus; and TA, tibialis anterior.

the most elastic energy (i.e., positive tendon work) among paretic and ipsilateral muscles for the limited community walker and control, respectively, although paretic SOL and GAS recovered less energy compared to the control (Fig. 4A). Because tendon is a passive tissue that uses little metabolic energy, the limited community walker's decreased ability to exploit elastic energy recovery via the paretic ankle plantar flexors would contribute to increasing their metabolic cost of walking compared to the speed-matched control. As a result, increased MT work by other paretic muscles occurred to compensate for paretic plantar flexor deficits. Specifically, increased negative work by paretic VAS and RF (Fig. 4A) suggests weakness of paretic SOL, which is consistent with a simulation analysis of potential compensatory mechanisms in nondisabled walking that found increased negative MT work by VAS and RF in response to SOL weakness (Goldberg and Neptune, 2007).

Similarly, increased positive work by paretic IL (Fig. 4A) likely compensated for weakness of paretic GAS (Goldberg and Neptune, 2007). Also, although SAR was not included in the 2D analysis by Goldberg and Neptune (2007), increased positive work by paretic SAR (Fig. 4A) may have compensated for paretic SOL and GAS weakness by contributing to the paretic hip flexor moment, a compensation previously observed in experimental studies of hemiparetic walking (Nadeau et al., 1999). Paretic IL and SAR generated positive work actively (i.e., positive work by the contractile element) incurring a metabolic cost, in contrast to the ipsilateral IL that generated much positive work via recovery of elastic energy in the tendon (Fig. 4A). Reduced paretic hip extension, which has been previously shown to limit propulsion generated by the paretic leg during pre-swing (Peterson et al., 2010b), also limited the recovery of elastic energy by the paretic hip flexors in the limited community walker.

Increased fiber work by non-paretic muscles by the limited community walker during paretic pre-swing would also increase the metabolic cost of walking. The non-paretic leg did more fiber work than the contralateral control leg, with the greatest difference in positive fiber work, likely to provide body support in response to paretic plantar flexor deficits. The plantar flexors contribute to body support and forward propulsion during pre-swing in nondisabled walking (McGowan et al., 2008; Neptune et al., 2001) and the non-paretic leg was shown not to compensate for decreased forward propulsion by the paretic leg (Peterson et al., 2010a). While non-paretic and contralateral GMAX and HAM did much positive work in the limited community walker and control, respectively, consistent with previous simulation analyses of nondisabled walking (Neptune et al., 2004), non-paretic SOL and GAS also did much positive work while active during paretic pre-swing with the non-paretic knee more flexed during paretic pre-swing (Fig. 4B). However, non-paretic SOL and GAS also stored much more elastic energy, which may allow for this energy to be recovered during another phase of the gait cycle.

4.2. Community walker: Increased work by hip abductors and adductors

For the community walker, paretic hip abductors and adductors did more work and paretic GAS did more positive fiber work while recovering less elastic energy compared to the control (Fig. 6A). Paretic GMED and GMIN fibers generated force while shortening early in pre-swing and then lengthened later in pre-swing at which time

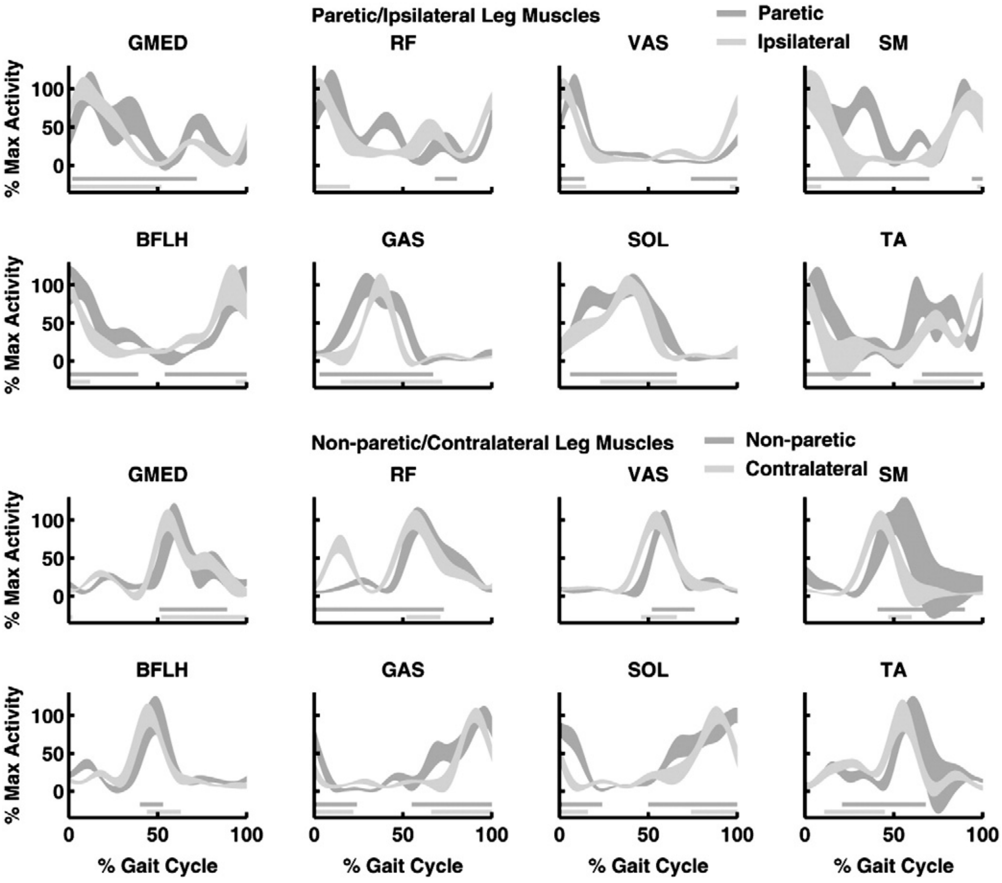

Fig. 2. For the community walker and control subject at 1.0 m/s, paretic (ipsilateral for control) and non-paretic (contralateral for control) EMG and optimized excitation timings (bars below the EMG indicate when a muscle is excited) normalized to the paretic leg (ipsilateral for control) gait cycle. Average EMG is shown with ± one standard deviation of the 30 s walking trial. Muscle excitations were optimized from mid-stance to toe-off corresponding to 35 to 74% of the paretic gait cycle for the community walker and 36 to 74% of the ipsilateral gait cycle for the control subject. The abbreviated muscle names are: GMED, gluteus medius; RF, rectus femoris; VAS, vastus lateralis; SM, semimembranosus; BFLH, biceps femoris long head; GAS, medial gastrocnemius; SOL, soleus; and TA, tibialis anterior.

AM fibers generated force while shortening. The QFEM fibers generated force while lengthening during much of pre-swing. Previously, it was shown that paretic GMED and AM had offsetting contributions to forward propulsion and swing initiation in the community walker and paretic GMIN and QFEM had only small contributions to swing initiation (Peterson et al., 2010a). Thus, the

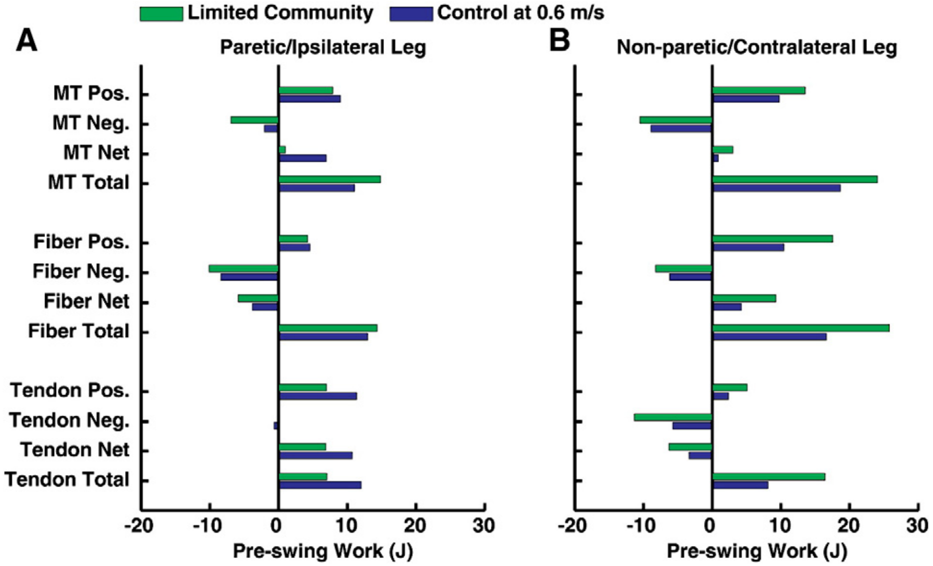

Fig. 3. For the limited community walker and speed-matched control, musculotendon (MT), muscle fiber, and tendon work by all paretic (ipsilateral for control) and non-paretic (contralateral for control) muscles during paretic (ipsilateral for control) pre-swing. (A) Paretic leg negative MT work was increased and positive tendon work was decreased relative to the ipsilateral control leg. (B) Net and total non-paretic MT work were increased compared to the contralateral control leg.

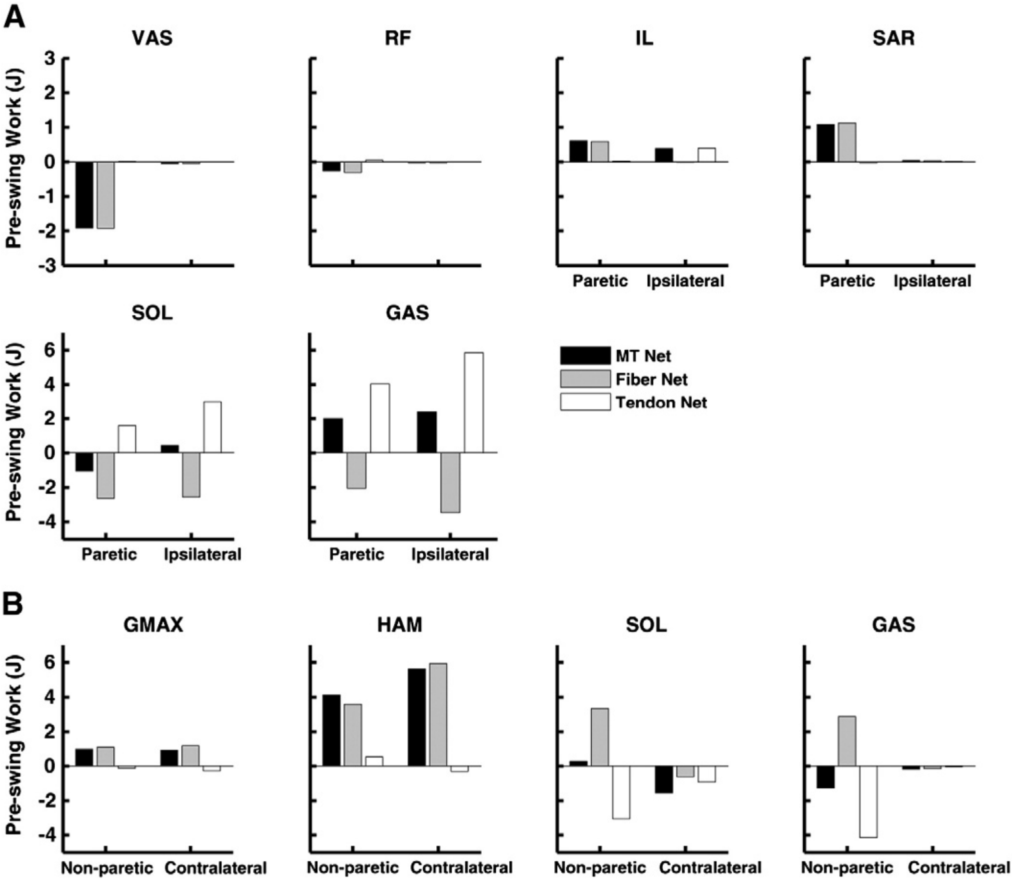

Fig. 4. Net musculotendon (MT), fiber and tendon work by individual muscles during paretic (ipsilateral for control) pre-swing for the limited community walker and control subject at 0.6 m/s. (A) Paretic VAS and RF had increased negative fiber work and paretic IL and SAR had increased positive fiber work compared to the ipsilateral control leg. Positive tendon work by paretic SOL and GAS were decreased relative to the ipsilateral control leg. (B) Non-paretic and contralateral GMAX and HAM did much positive work in the limited community walker and control, respectively. Non-paretic SOL and GAS had increased positive fiber work and negative tendon work compared to the contralateral control leg. See Table 1 for muscle group abbreviations.

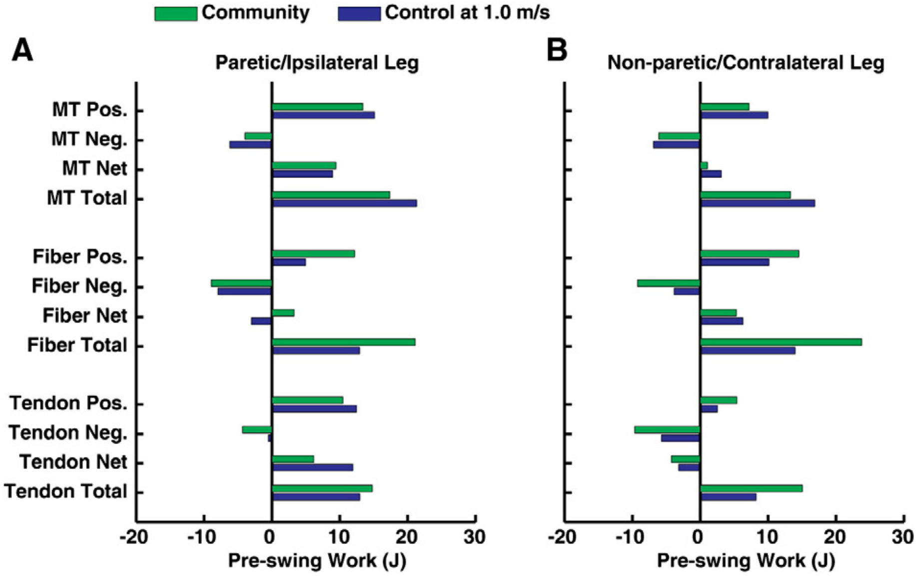

Fig. 5. For the community walker and speed-matched control, musculotendon (MT), muscle fiber, and tendon work by all paretic (ipsilateral for control) and non-paretic (contralateral for control) muscles during paretic (ipsilateral for control) pre-swing. (A) Paretic leg positive fiber work and negative tendon work were increased relative to the ipsilateral control leg. (B) Net non-paretic fiber and tendon work were similar to the contralateral control leg, but total non-paretic fiber and tendon work were increased relative to the contralateral leg.

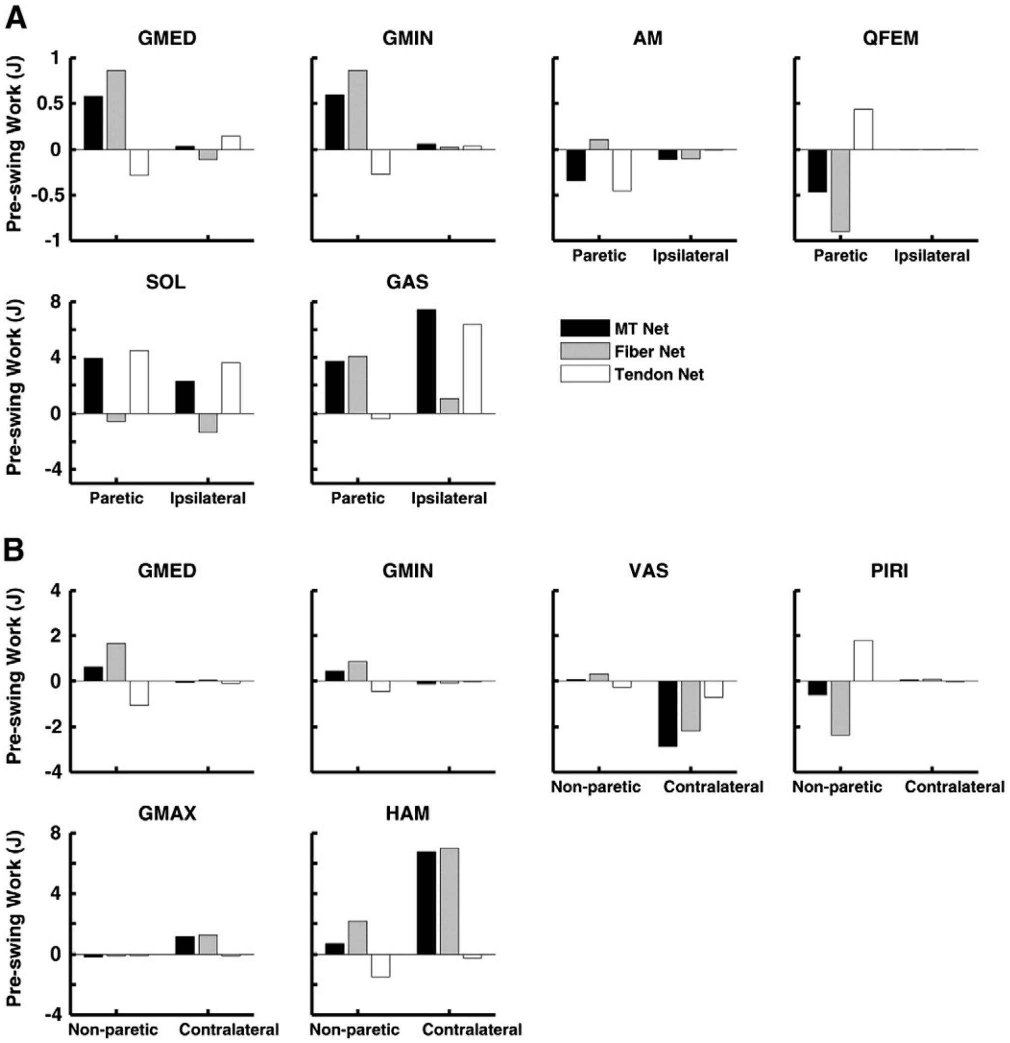

Fig. 6. Net musculotendon (MT), fiber and tendon work by individual muscles during paretic (ipsilateral for control) pre-swing for the community walker and control subject at 1.0 m/s. (A) Positive fiber work was increased and positive tendon work was decreased in paretic GAS compared to the ipsilateral control leg. Paretic GMED, GMIN, AM and QFEM generated more fiber and tendon work in the community walker compared to the ipsilateral control leg. (B) Fiber work by non-paretic GMED, GMIN and PIRI was increased and fiber work by non-paretic VAS, GMAX, and HAM was decreased in the community walker relative to the contralateral control leg. See Table 1 for muscle group abbreviations.

increased work performed by these muscles compared to the control did not contribute to an increased walking speed. Although paretic GAS did contribute to an increased walking speed (i.e., previously found to contribute to forward propulsion and swing initiation; Peterson et al., 2010a), increased positive fiber work by paretic GAS was necessary since it recovered less elastic energy, which would contribute to increase metabolic cost compared to the control.

Also for the community walker, non-paretic hip abductors did more work compared to the contralateral control leg during pre-swing. Instead of positive work by GMAX as seen in the contralateral leg, non-paretic GMED and GMIN did much positive fiber work in addition to HAM (Fig. 6B). PIRI did the most negative fiber work of all non-paretic muscles in the community walker, while VAS did the most negative fiber work of all contralateral muscles in the control (Fig. 6B), which is in agreement with previous simulation analyses of nondisabled walking (Neptune et al., 2004). Non-paretic hip abductors did more fiber work compared to the contralateral leg likely to enhance stability during paretic pre-swing (Corriveau et al., 2004).

4.3. Limitations

Musculotendon parameters based on nondisabled subjects were used to model both the control and hemiparetic subjects, although paretic and non-paretic fiber and tendon stiffness may be altered post-stroke (Svantesson et al., 2000). However, the mechanical work quantities are relatively insensitive to model parameters since they are derived from musculotendon forces and velocities, which are robust for a given movement. For example, the resulting simulation musculotendon forces are largely dictated by the dynamics of the task (i.e., specific levels of force are required to satisfy the mechanical energetics). Thus, if the musculoskeletal model has inaccurate parameters for a given subject (e.g., too high or low maximum isometric force), the optimization algorithm would simply scale the muscle excitation magnitude to compensate and the muscle force remains at the level necessary to satisfy the experimental walking dynamics. Similarly, the musculotendon velocities largely depend on the body segment velocities, which compare well with the experimental data. Thus, findings such as the decreased tendon work by paretic SOL and GAS of the limited community walker and by paretic GAS of the community walker being due to decreased force generated by these muscles would not differ with changes in muscle and tendon parameters.

Another limitation was that metabolic cost was inferred from muscle mechanical work. However, a recent simulation study of nondisabled walking that included a metabolic cost model (Umberger, 2010) found that the metabolic energy expended compared well with experimental measures and was consistent with muscle mechanical work during each region of the gait cycle (Neptune et al., 2004). Also, this study was limited to the pre-swing (double support) phase and differences in mechanical work production between hemiparetic and nondisabled walkers likely exist during other phases of the gait cycle. Although the metabolic power during pre-swing is low compared to other phases of the gait cycle (Umberger, 2010), the decreased cost during pre-swing is attributed to the recovery of elastic energy by the plantar flexors, which the present study found to be an important deficit in the hemiparetic walkers. Nonetheless, quantifying mechanical work production during other phases of the gait cycle and using metabolic cost models to understand the relationships between muscle mechanical work and efficiency are important areas of future work to provide insight into the increased metabolic cost of hemiparetic walking. Finally, although similar mechanical work values are expected from subjects walking with similar kinematics and kinetics as the two representative hemiparetic subjects, the hemiparetic population is heterogeneous such that analyses of more subjects are needed to fully understand the relationships between energy expenditure and functional walking status in hemiparetic subjects.

5. Conclusions

Total paretic and non-paretic fiber work were increased in both the limited community and community hemiparetic walkers compared to an age-matched control subject walking at similar speeds. Increased fiber work in the limited community walker was primarily related to decreased fiber and tendon work by the paretic plantar flexors requiring compensatory work by other muscles. Increased fiber work in the community walker was primarily related to increased work by the hip abductors and adductors. Thus, if the hemiparetic and control subjects were to perform work with the same mechanical efficiency, the hemiparetic walkers would expend more metabolic energy during pre-swing. These results may partly explain why hemiparetic walkers have an increased metabolic cost compared to nondisabled walkers at matched speeds and suggest that post-stroke rehabilitation strategies that promote walking kinematics and kinetics similar to nondisabled walkers may be beneficial for decreasing the metabolic cost. Future work using detailed models of muscle metabolic cost to analyze the entire gait cycle will provide further insight into the relationships between muscle mechanical work and efficiency in hemiparetic walking.

Conflict of interest statement

None.

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References

Anderson, F.C., 1999. A dynamic optimization solution for a complete cycle of normal gait. Doctoral Dissertation, The University of Texas at Austin.

Audu, M.L., Davy, D.T., 1985. The influence of muscle model complexity in musculoskeletal motion modeling. J. Biomech. Eng. 107, 147–157.

Biewener, A.A., Farley, C.T., Roberts, T.J., Temaner, M., 2004. Muscle mechanical advantage of human walking and running: implications for energy cost. J. Appl. Physiol. 97, 2266–2274.

Bowden, M.G., Balasubramanian, C.K., Neptune, R.R., Kautz, S.A., 2006. Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking. Stroke 37, 872–876.

Chen, G., Patten, C., 2008. Joint moment work during the stance-to-swing transition in hemiparetic subjects. J. Biomech. 41, 877–883.

Chen, G., Patten, C., Kothari, D.H., Zajac, F.E., 2005. Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds. Gait Posture 22, 51–56.

Corriveau, H., Hebert, R., Raiche, M., Prince, F., 2004. Evaluation of postural stability in the elderly with stroke. Arch. Phys. Med. Rehabil. 85, 1095–1101.

De Quervain, I.A., Simon, S.R., Leurgans, S., Pease, W.S., Mcallister, D., 1996. Gait pattern in the early recovery period after stroke. J. Bone Joint Surg. Am. 78, 1506–1514.

Delp, S.L., Loan, J.P., Hoy, M.G., Zajac, F.E., Topp, E.L., Rosen, J.M., 1990. An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. IEEE Trans. Biomed. Eng. 37, 757–767.

Den Otter, A.R., Geurts, A.C., Mulder, T., Duysens, J., 2004. Speed related changes in muscle activity from normal to very slow walking speeds. Gait Posture 19, 270–278.

Den Otter, A.R., Geurts, A.C., Mulder, T., Duysens, J., 2007. Abnormalities in the temporal patterning of lower extremity muscle activity in hemiparetic gait. Gait Posture 25, 342–352.

Detrembleur, C., Dierick, F., Stoquart, G., Chantraine, F., Lejeune, T., 2003. Energy cost, mechanical work, and efficiency of hemiparetic walking. Gait Posture 18, 47–55.

Goffe, W.L., Ferrier, G.D., Rogers, J., 1994. Global optimization of statistical functions with simulated annealing. J. Econometrics 60, 65–99.

Goldberg, E.J., Neptune, R.R., 2007. Compensatory strategies during normal walking in response to muscle weakness and increased hip joint stiffness. Gait Posture 25, 360–367.

Ishikawa, M., Komi, P.V., Grey, M.J., Lepola, V., Bruggemann, G.P., 2005. Muscle-tendon interaction and elastic energy usage in human walking. J. Appl. Physiol. 99, 603–608.

Kautz, S.A., Neptune, R.R., 2002. Biomechanical determinants of pedaling energetics: internal and external work are not independent. Exerc. Sport Sci. Rev. 30, 159–165.

Knutsson, E., Richards, C., 1979. Different types of disturbed motor control in gait of hemiparetic patients. Brain 102, 405–430.

Mcgowan, C.P., Neptune, R.R., Kram, R., 2008. Independent effects of weight and mass on plantar flexor activity during walking: implications for their contributions to body support and forward propulsion. J. Appl. Physiol. 105, 486–494.

Menegaldo, L.L., De Toledo Fleury, A., Weber, H.I., 2004. Moment arms and musculotendon lengths estimation for a three-dimensional lower-limb model. J. Biomech. 37, 1447–1453.

Nadeau, S., Gravel, D., Arsenault, A.B., Bourbonnais, D., 1999. Plantarflexor weakness as a limiting factor of gait speed in stroke subjects and the compensating role of hip flexors. Clin. Biomech. 14, 125–135.

Neptune, R.R., Wright, I.C., Van Den Bogert, A.J., 2000. A method for numerical simulation of single limb ground contact events: application to heel-toe running. Comput. Methods Biomech. Biomed. Engin. 3, 321–334.

Neptune, R.R., Kautz, S.A., Zajac, F.E., 2001. Contributions of the individual ankle plantar flexors to support, forward progression and swing initiation during walking. J. Biomech. 34, 1387–1398.

Neptune, R.R., Zajac, F.E., Kautz, S.A., 2004. Muscle mechanical work requirements during normal walking: the energetic cost of raising the body's center-of-mass is significant. J. Biomech. 37, 817–825.

Neptune, R.R., Sasaki, K., Kautz, S.A., 2008. The effect of walking speed on muscle function and mechanical energetics. Gait Posture 28, 135–143.

Niam, S., Cheung, W., Sullivan, P.E., Kent, S., Gu, X., 1999. Balance and physical impairments after stroke. Arch. Phys. Med. Rehabil. 80, 1227–1233.

Olney, S.J., Griffin, M.P., Monga, T.N., Mcbride, I.D., 1991. Work and power in gait of stroke patients. Arch. Phys. Med. Rehabil. 72, 309–314.

Perry, J., 1993. Determinants of muscle function in the spastic lower extremity. Clin. Orthop. Relat. Res. 10–26.

Perry, J., Garrett, M., Gronley, J.K., Mulroy, S.J., 1995. Classification of walking handicap in the stroke population. Stroke 26, 982–989.

Peterson, C.L., Hall, A.L., Kautz, S.A., Neptune, R.R., 2010a. Pre-swing deficits in forward propulsion, swing initiation and power generation by individual muscles during hemiparetic walking. J. Biomech. 43, 2348–2355.

Peterson, C.L., Cheng, J., Kautz, S.A., Neptune, R.R., 2010b. Leg extension is an important predictor of paretic leg propulsion in hemiparetic walking. Gait Posture 32, 451–456.

Raasch, C.C., Zajac, F.E., Ma, B., Levine, W.S., 1997. Muscle coordination of maximum-speed pedaling. J. Biomech. 30, 595–602.

Sasaki, K., Neptune, R.R., Kautz, S.A., 2009. The relationships between muscle, external, internal and joint mechanical work during normal walking. J. Exp. Biol. 212, 738–744.

Stoquart, G.G., Detrembleur, C., Nielens, H., Lejeune, T.M., 2005. Efficiency of work production by spastic muscles. Gait Posture 22, 331–337.

Svantesson, U., Takahashi, H., Carlsson, U., Danielsson, A., Sunnerhagen, K.S., 2000. Muscle and tendon stiffness in patients with upper motor neuron lesion following a stroke. Eur. J. Appl. Physiol. 82, 275–279.

Umberger, B.R., 2010. Stance and swing phase costs in human walking. J. R. Soc. Interface 7, 1329–1340.

Winters, J.M., Stark, L., 1988. Estimated mechanical properties of synergistic muscles involved in movements of a variety of human joints. J. Biomech. 21, 1027–1041.

Zajac, F.E., 1989. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. Crit. Rev. Biomed. Eng. 17, 359–411.

Zamparo, P., Francescato, M.P., De Luca, G., Lovati, L., Di Prampero, P.E., 1995. The energy cost of level walking in patients with hemiplegia. Scand. J. Med. Sci. Sports 5, 348–352.