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Effects of stimulated hip extension moment and position on upper-limb support forces during FNS-induced standing—A technical note

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Abstract—This study explores the effects of active hip extension moment produced by electrical stimulation on the support forces the arms must exert through an assistive device during quiet erect standing with functional neuromuscular stimulation (FNS) in individuals with spinal cord injuries (SCI). A static sagittal plane biomechanical model of human standing was developed to predict the effects of stimulated hip extension moment and sagittal plane hip angle on the arm support necessary to maintain an upright posture. Two individuals with complete thoracic SCI were then tested while they stood with continuous stimulation to the knee and trunk extensors. The steady-state active extension moment exerted at the hip was varied by activating different combinations of hip extensor muscles with continuous stimulation while steady-state support forces applied to the arms and feet during standing were measured. The steady-state support forces imposed on the arms during quiet standing decrease with increased stimulated hip extension moment and are highly dependent upon hip flexion angle, as predicted by the biomechanical simulations. Experimentally, the combination of gluteus maximus and semimembranosus stimulation produced three times more steady-state hip extension moment than did stimulation of the gluteus maximus and adductor magnus. This resulted in a ten-fold decrease in body weight supported on the arms. More vertical postures (smaller hip flexion angles) improve the effectiveness of the hip extensor muscles in reducing the support forces placed on the arms. A single Newton-meter of stimulated hip extension moment with the hips fixed at 5º of flexion results in almost five times the reduction in arm support forces as with the hips at 20º. To minimize the forces applied by the arms on an assistive device for support while standing with FNS, these preliminary results suggest that (1) efforts should be made to assume the most erect postures possible and (2) muscles and stimulation paradigms that maxiize active hip extension moment should be chosen.

Key words: *FNS, functional neuromuscular stimulation, posture, spinal cord injury.*

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INTRODUCTION

The purpose of this study was to gain insight into the relationship between the magnitude of sagittal plane hip extension moment generated with electrical stimulation and the vertical support forces imposed on an assistive device by the upper limbs with the body standing in as erect a posture as possible with functional neuromuscular stimulation (FNS).

Preliminary clinical trials of lower-limb neuroprostheses for standing clearly indicate that continuous open-loop stimulation of the trunk, hip, and knee extensors can allow people with paraplegia to overcome physical obstacles (1), negotiate architectural barriers (2), and exert a greater control over their environment by affording them the ability to reach and manipulate objects that are otherwise inaccessible from the wheelchair (3,4). Standing with FNS has been achieved with relatively simple systems consisting of two to six channels of continuous surface stimulation (5–7), and such multichannel surface stimulation systems have successfully provided standing and stepping movements to people with spinal cord injuries (SCI) in both laboratory and clinical settings (8–10). Lower-limb FNS systems employing intramuscular electrodes with percutaneous leads have also been successful in providing lower-limb functions to individuals with paraplegia (11,12). Subjects using 16 or fewer channels of percutaneous intramuscular stimulation can perform simple mobility and one-handed reaching tasks while standing (2,13). More recently, totally implanted pacemaker-like neuroprostheses for standing after SCI have undergone feasibility and initial clinical testing. Exercise and standing have been achieved with the use of an implanted system consisting of a modified cochlear prosthesis that was reconfigured to deliver 22 channels of stimulation to the nerves of the major lower-limb muscles (14). Multichannel stimulators for activating the L2–S2 motor roots for lower-limb function have also been applied to a handful of volunteers (15). In our laboratory, implantable systems for standing have been undergoing successful clinical testing since 1992. **Figure 1** shows an individual with complete motor and sensory mid-thoracic paraplegia using an implanted standing system consisting of an eight-channel receiver-stimulator (16), and epimysial (17) and surgically implanted intramuscular (18) electrodes.

Regardless of the technology used to deliver stimulation, all lower-limb FNS systems still require assistive devices, such as crutches or walkers, to allow the upper limbs to inject the corrective forces necessary to maintain **balance**. The magnitudes of these corrective forces can be quite small (on the order of 10 percent of body weight or less) (19) and can be produced by a single limb without undue exertion (20), freeing the other hand to perform reaching tasks or other functional activities. However, standing performance can vary greatly from individual to individual, and not every result is as good as that pictured

Figure 1.
Standing with continuous stimulation to trunk, hip, and knee extensors. Balance must be maintained by one limb on an assistive device.

in **Figure 1**. Frequently both upper limbs must be used to provide large supplemental forces to support the body against collapse, in addition to minor corrective forces for balance. In this case, both limbs are needed to maintain an erect posture through an assistive device, thus compromising the utility of FNS systems. This is commonly observed during prolonged standing with FNS. As the hip extensor muscles fatigue with continuous stimulation, more weight is placed on the arms to remain upright as the trunk and pelvis rotate forward in the sagittal plane although users easily maintain medial-lateral balance. This paper assumes that the corrective forces required to maintain dynamic **balance** are small and easily produced and focuses instead on the factors that contribute to the need to use the arms for static **support**.

Without the advantage of external mechanical orthoses, use of the arms to support the body weight against gravity during FNS-induced standing is necessitated by three primary factors: (1) inadequate stimulated hip extension moment, (2) inability to assume an erect and mechanically stable posture, and (3) lack of control of the flexible paralyzed trunk. If the trunk can be stabilized through stimulation of the paraspinal muscles, adequate stimulated hip extension moment can reduce the need to use the upper limbs by preventing forward bending, thus maintaining a more neutral postural alignment. While large stimulated extension moments and full ranges of motion are routinely observed at the knee, achieving adequate active extension moment and a fully extended posture at the hip is often difficult for a variety of reasons including (1) less than full recruitment of the hip extensor muscles, (2) common range of motion limitations and flexion contractures at the hip joint, and (3) inadvertent activation of the hip flexor muscles (rectus femoris, sartorius, iliopsoas). The latter refers to the phenomenon commonly referred to as “spillover” in which responses from undesired structures are recruited as stimulation levels to the primary target nerve and muscle increase. In this case, electrodes placed on or near the femoral nerve to recruit the vasti of the quadriceps to extend the knee can also spill over to muscles that actively flex the hip and compromise erect standing posture.

The purpose of this investigation was to explore the relationships between stimulated hip extension moment, sagittal plane hip angle, and the vertical support forces applied through an assistive device by the arms during quiet FNS-induced standing by individuals with SCI. A simulation study was performed to gain insight into the interactions between the biomechanical factors, followed by case studies involving two individuals with SCI standing with FNS. To focus the investigation on the interdependence of active hip extension, hip angle, and arm support forces, we stiffened the flexible lumbar spine with electrical stimulation so that the pelvis and trunk acted together as a single unit. Understanding the relationship between arm support forces, stimulated hip extension moment, and hip angle should advance the goal of designing FNS systems that will routinely allow the user to release one hand from an assistive device to manipulate objects in the surrounding environment.

METHODS

Biomechanical Simulation Studies: A static sagittal plane model of human standing was developed to estimate the steady-state hip extension moments and vertical arm support forces required to maintain a constant upright standing posture and to understand how changes in hip extension moment affect the arm support forces at a given hip flexion angle. The static assumption implies that the moments required to maintain upright posture at a certain fixed position are determined only by gravitational forces. Standing with FNS is achieved with continuous stimulation to the knee, hip, and trunk muscles to fix the position of the joints and prevent collapse (i.e., resist gravitational forces). The normal postural control systems to maintain balance are absent and replaced only with constant activation of the paralyzed musculature and interactions with an assistive device. With an individual in the quiet standing position with both hands on a walker or parallel bars, variations in the center of mass due to postural sway can be expected to be quite small and occur very slowly in the absence of external disturbances. The slowly varying and limited excursion of the body’s mass center and lack of other perturbations during quiet upright standing in this manner imply that any dynamic contributions to the steady-state joint moments can be reasonably ignored. Since the goal of this study was to investigate only the requirements to support the body against collapse, rather than to study balance or quantify stability in terms of sway, the static assumption was well justified.

Restricting the model to the sagittal plane is a simplifying assumption that is justified by the symmetrical standing position under study and the general degree of symmetry observed in the lower limbs. Previous analyses of the lower limbs of individuals with SCI indicate an almost 80 percent agreement between the right and left sides in terms of anthropomorphic properties and stimulated responses (21). The standing posture of concern in this investigation is symmetrical in the coronal plane and therefore conducive to a sagittal plane analysis. Static standing with FNS in this study located the feet shoulder width apart and in line with the pelvis (coincident in the coronal plane) and with each hand located the same distance from the body at a fixed position on a set of parallel bars. Because the principal variables of concern were total stimulated hip extension moment and total upper-limb support force, medial-lateral shifting, lateral bending, and axial rotation were minimized in this static steady-state standing posture. Consequently, it was reasonable to assume that the forces exerted by gravitational reactions of trunk, head, and upper-limb segments would be transmitted equally to each leg and handle of the support device. Similar sagittal plane models have been employed extensively in the past to study the mechanics of human stance (22–24).

The body was modeled as six rigid body segments acting only in the saggital plane connected by frictionless hinge joints (see **Figure 2**). A coordinate system was attached to the proximal end of each upper-limb segment and to the distal end of the lower-limb and trunk segments. The transformation from one reference frame to another frame was accomplished by a set of translational and rotational matrices, which depended on the segment length and were described in terms of the position of the connecting segment. The location (25) and the magnitudes (26) of mass centers for each segment were assigned according to nominal values published for an able-bodied male of average stature (1.80 m tall, 73.42 kg). Application of external support forces was constrained to specific locations corresponding to the foot center of mass (lower-limb support force) and to the wrist joint (upper-limb support force).

Simulations to investigate the relationship between hip extension moment, hip angle, and support forces were performed with the model in a typical posture assumed while standing with FNS (knees fully extended, ankles in neutral position, and arms directed downward). The moments required at the hip (without arm support) to maintain upright posture were calculated as the hip angle ranged from 5º to 25º of hip flexion. Similarly, the upper-limb forces required to maintain an erect posture (both in the absence of active hip extension and at various values of applied hip extension moment) were also estimated and the reduction in support forces exerted by the arms due to increasing hip extension moment were computed in simulation. Results were normalized to the nominal body weight of 73.42 kg for ease of analysis.

Laboratory Experiments: The interactions between stimulated hip extension moment, hip flexion angle, and upper/lower-limb support forces were determined experimentally with two well-conditioned individuals with

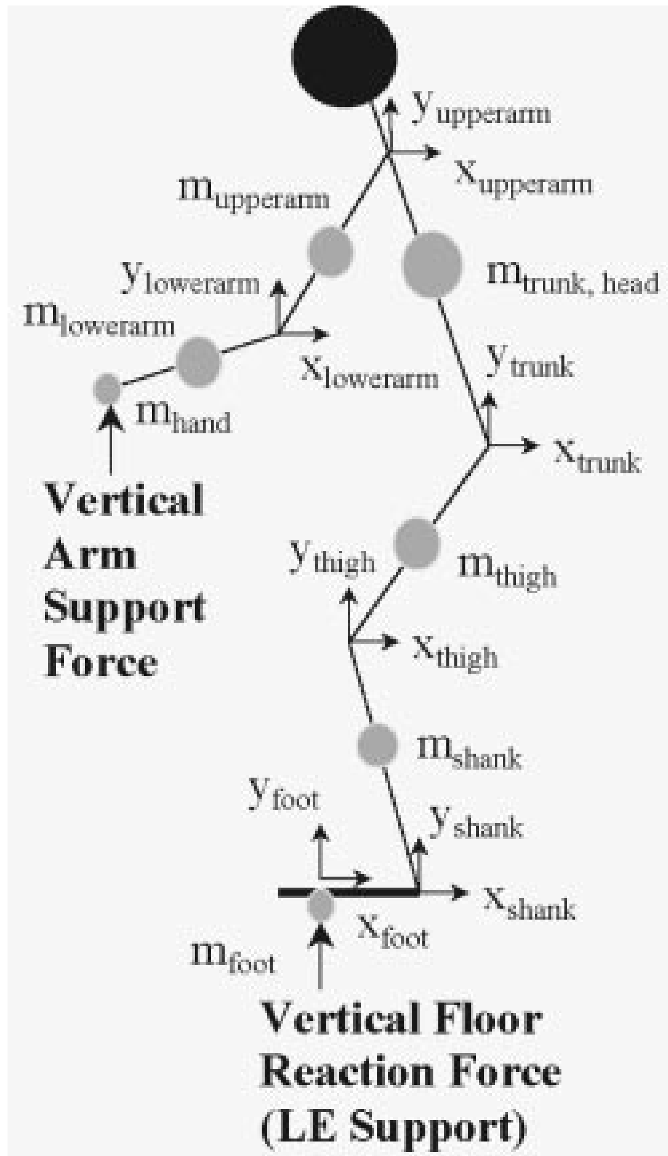

Figure 2.
A static 2D model of human skeletal system in sagittal plane developed based on body parameters of a 1.80-m, 73.42-kg healthy, able-bodied male. Model consisted of six rigid-body segments, which are connected by frictionless hinge joints. Forces exerted by gravitational reactions of trunk, head, and upper-limb segments are assumed to be transmitted equally to each leg.

thoracic SCI. Each subject was a long-term user of chronically indwelling intramuscular electrodes with percutaneous leads (27,28) or surgically implanted epimysial electrodes and an implanted receiver-stimulator (29). Each volunteer had participated in other clinical and laboratory studies of exercise and standing with FNS. The physical characteristics of the research subjects are summarized in **Table 1.**

The minimal muscle set selected for standing included one channel of trunk extensors (erector spinae), one channel of knee extensors (quadriceps), and one to three channels of hip extensor muscles (gluteus maximus, posterior portion of adductor magnus, and semimembranosus), bilaterally. Prior to the experiment, each implanted muscle was profiled to determine its characteristics in response to a 20-mA charge-balanced asymmetric biphasic stimulus pulse train at 33 pulses per second (pps). Stimulated responses were rated in terms of threshold (below which no contraction was observed) and saturation (above which no further increase in strength was observed, or spillover to unwanted muscles or reflex activation occurred) values of pulse duration. Stimulation during further testing was limited to saturation values to ensure selective activation and isolated recruitment of the primary muscles of interest. The active moment generating capacities of the hip extensor muscles were determined prior to the experiment on an instrumented CYBEX II® dynamometer (30) (Lumex Inc., Ronkonkoma, NY). To best simulate the active moments possible with FNS while standing, subjects were secured on the dynamometer in the supine position with the knee locked in extension by an external brace. Steady-state hip extension moment was measured isometrically (with the hip close to full extension) in response to 2-second bursts of stimulation separated by 10 seconds of rest to minimize the effects of fatigue. The gluteus maximus, adductor magnus, and semimembranosus muscles were activated individually and in all possible two- and three-muscle combinations in random order until at least ten repetitions under each condition were recorded. This procedure is illustrated graphically in **Figure 3**, which shows a representative stimulation sequence and dynamometer output. The results of these calibration trials were later used to vary the stimulated hip extension moment during standing by activating different muscles or combinations of musces as described below.

For standing trials, both feet were placed on the same biomechanics platform (AMTI® Biomechanics, Newton, MA) to measure the total vertical support force on the lower limbs. Subjects placed their hands at fixed locations on a set of parallel bars instrumented with strain gauges (31,32) to measure the support forces on the arms during standing (see **Figure 4**). The forces from left and right sides were added to yield the total upper-limb support force. This procedure was adopted so that the measurements of upper- and lower-limb support forces would conform to the sagittal plane biomechanical model previously described.

During the sit-to-stand transition, which lasted for approximately 1.5 seconds, the pulse durations of all electrodes were linearly increased from zero to the saturation values. The design of the simulation patterns used during this transition was based on that reported by Bajd et al. (33), in which the order that muscles were recruited during rising was derived from the EMG patterns of able-bodied individuals.

Table 1.
Physical characteristics of two volunteers for experimental determination of relationships between hip extension, posture and upper-limb support forces. Subjects exhibited neurologigally stable motor and sensory complete SCI, and were well-conditioned and experienced in standing with FNS.

Subject	Sex	Age	Height (in.)	Weight (lb)	Injury level	Injury date	FNS system configuration	FNS experience
A	M	45	69	202	T7 Complete	3/91	Intramuscular electrodes with percutaneous leads to external stimulator	Exercise and standing with FNS since 5/93
B	M	40	60	160	T10 Complete	5/91	Epimysial electrodes with implanted receiver-stimulator	Exercise, standing and walking with FNS since 11/96

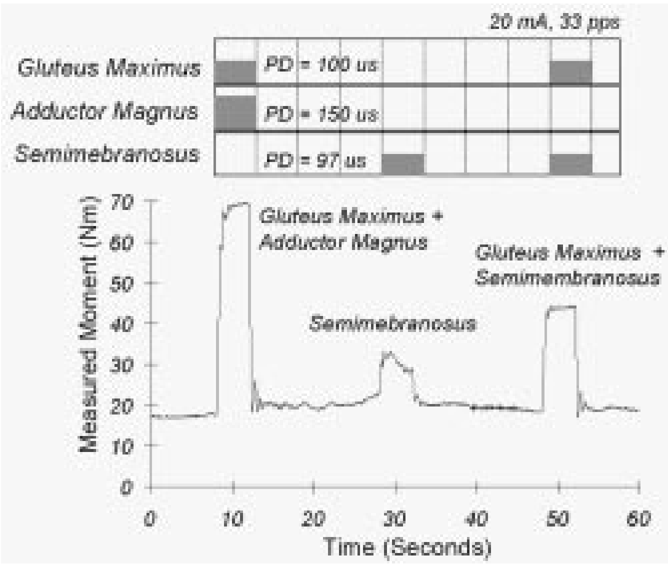

Figure 3.
Sample stimulus pattern and dynamometer output for calibrating steady-state active isometric moment-generating capacities of hip extensor muscles (individually and in all possible combinations). Pulse durations (PD) were set to maximal values without unwanted reflex activity or spillover to other muscles. Order was randomized and sufficient rest intervals were allowed to minimize fatigue.

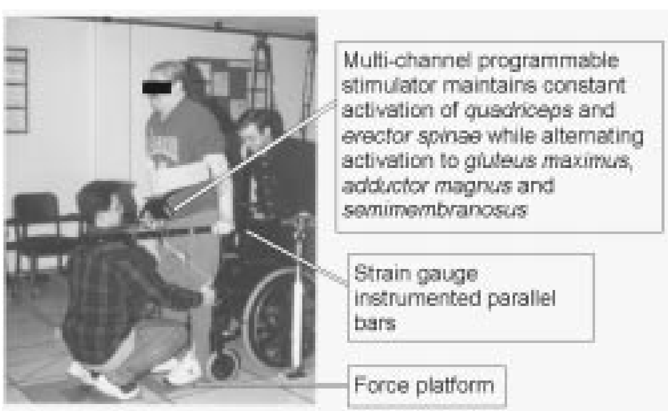

Figure 4.
Experimental setup of Subject A, who stood on a force platform between two strain-gauge instrumented paralled bars. Erector spinae, quadripeps, and different hip muscles were activated with a constant 20-mA charge-balanced asymmetrical biphasic waveform. Hip flexion angle during each standing trial was measured by a goniometer.

After subjects achieved a steady upright posture with all hip extensors active, the hip extensor muscles of one leg were alternately turned off and on for a fixed interval in random order until five "off" periods and six "on" periods were obtained in a single standing trial. The interval was set to approximately 3 percent of the total standing duration for each subject to minimize the effects of fatigue. Subjects underwent five standing trials on each of at least 2 days. Stimulation patterns for each trial consisted of all extensor muscles turned off and each single extensor muscle turned off one at a time while maintaining stimulation of the others. The order of deactivation was randomized to reduce the effective duty cycle of each hip extensor muscle and to distribute the effects of fatigue equally across all combinations. The forces on the arms and feet were sampled at 100 Hz, and the hip flexion angle was monitored throughout the procedure by a goniometer.

RESULTS

The upper panel in **Figure 5** shows the initial part of the stimulation pattern used with Subject A. The lower panel illustrates the variation in support forces as stimulation to the extensor muscles (and therefore applied active hip extension moment) varied throughout the experimental procedure. The data represent the support forces recorded as the gluteus maximus was turned off and then reapplied while maintaining activation of all

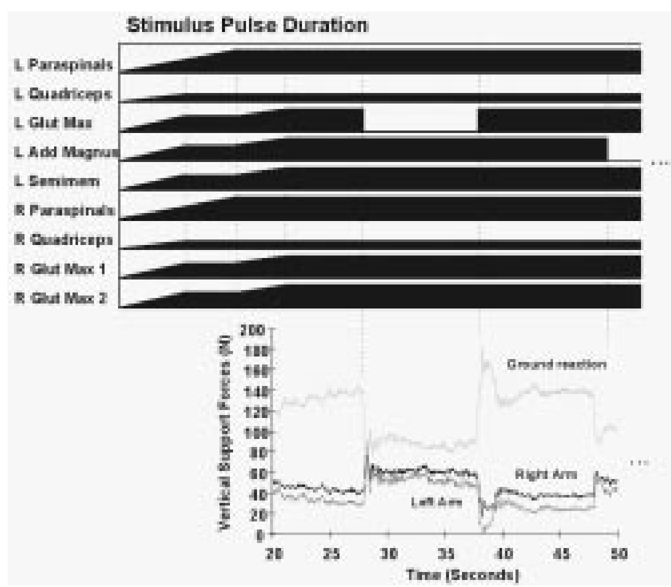

Figure 5.
Initial seconds of stimulation pattern used to induce standing in Subject A (top) and corresponding measured upper- and lower-limb support forces (bottom). Active hip extension moment was varied by changing the selection of hip extensor muscles activated, while support forces were measured via a force platform and instrumented paralled bars.

other hip extensors. We averaged the forces during the last half of each stimulation interval to ensure that steady state responses were achieved.

The total of the average forces (i.e., sum of upper- and lower-limb forces) was found to be identical to the actual body weights of each subject as measured by a standard scale. We computed the change in arm support due to deactivation of a single hip extensor muscle (or activation of a two hip extensor muscle combination) by subtracting the baseline arm forces obtained while standing with no active hip extensor muscles. The changes in arm forces were then normalized by the body weight of each subject.

The standard deviations of the support forces across trials during a single session were found to be small, ensuring intraday repeatability. Interday repeatability was assessed by comparing the measurements taken on different days, which were found not to differ significantly ($\alpha > 0.05$). Therefore, the records of upper- and lower-limb support forces acquired on test sessions conducted on different days were pooled for final analysis.

Figure 6 summarizes the upper-limb support force and hip extension moment required to maintain standing at various hip angles as predicted by the biomechanical simulations. The dashed gray curve shows the support force required by each arm in the absence of hip extension, while the solid black curve shows the hip extension moment required from each leg in the absence of any upper-limb contribution. The stimulated hip extension moment needed to remain upright was found to increase in proportion to hip flexion angle. A body posture of 5º of hip flexion required less than 1 N•m of hip extension moment to remain upright. For a hip angle of 25º, the required moment increased to 24 N•m. However, the upper-limb support forces required to remain upright increase very rapidly as the hip flexes. At a hip flexion angle of 5º (without any hip extension moment), each upper-limb must exert a unilateral force of 25 N to maintain an upright posture, while 100 N is required at 25º of flexion.

Figure 7 shows the model-predicted decreases in required upper-limb support forces as hip extension moment increases. Each of the dashed lines indicates the theoretical relationship between total upper-limb support force and hip extension moment at a given angle of hip flexion. Since the upward forces applied at the wrist joint create an extension moment around the hip, increasing the applied hip extension moment linearly decreased the arm forces needed to remain in an upright position. The rate at which the arm forces decrease varies as a function of hip flexion angle. The rates for different postures (assuming a 73.42-kg, 180-cm individual) are 50.9 N/N•m, 21.9 N/N•m, 14.1 N/N•m, 10.4 N/N•m, and 8.3 N/N•m for hip angles of 5º, 10º, 15º, 20º, and 25º, respectively. Stimulated hip extension moment is therefore more effective at reducing upper-limb support forces at more erect postures. That is, larger reductions in

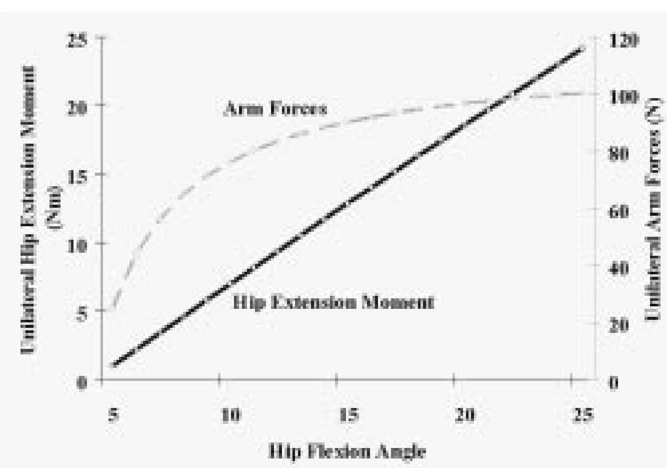

Figure 6.
Simulated hip extension moment in the absence of arm forces transmitted through an assistive device (solid) and simulated arm forces in absence of hip extensor actuators (dashed), as a function of hip flexion angle for maintenance of static upright posture.

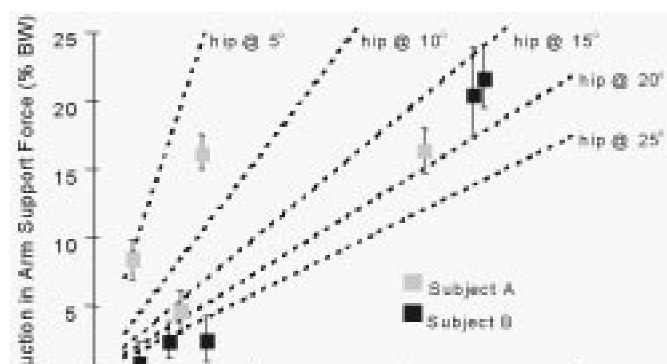

Figure 7.
Model-predicted (dashed lines) and experimentally measured (squares) reduction of upper-limb support forces as a function of changes in active hip extension moment for different hip joint angles. Increased active hip extension moment or more erect postures are effective means to decrease upper-limb support forces. Stimulated hip extensors are more effective in more erect postures.

upper-limb exertion will be achieved in more erect postures for the same stimulated hip extension moment. Conversely, the rate of arm support reduction with applied hip moment is lower with increasing hip flexion angle. Therefore, more hip extension moment would be required to make a small reduction of arm forces with the hip flexed than in an erect standing posture.

Experimental measurements during standing with FNS were consistent with the simulation results. **Table 2** summarizes the means and standard deviations of support forces measured from Subjects A and B while standing with different combinations of muscles (and therefore, different applied hip extension moments). The measured support forces during standing with any two-muscle combination is bounded by the measured support forces during standing with all three hip extensor muscles (lower bound) and the measured arm forces during standing without any hip extensor muscles (upper bound).

The experimentally measured reduction of arm forces as a function of the change in hip extension moment is also shown in **Figure 7**. The gray squares represent data obtained from Subject A, while the black squares represent the upper-limb support force and moment measurements collected from Subject B. The hip flexion angle of each subject during standing was fixed regardless of muscles activated or the applied hip extension moments. Based on goniometric measurements, Subject A was observed to stand between 10º to 15º of hip flexion angle. Subject B was observed to stand between 15º to 20º of hip flexion. From **Figure 7**, it is evident that the experimental data for Subjects A and B coincide with the theoretically expected reduction in upper-limb support based on their self-selected standing postures.

In both subjects, larger reductions of arm forces were found when the applied moments at the hip increased. Stimulated hip muscles that produce the highest extension moment yielded standing with the largest reduction of arm support forces. The combination of gluteus maximus and adductor magnus that generated an increase of 2.4 N•m of isometric hip extension moment resulted in a reduction of arm forces by only 2.5 percent of the subject body weight. In contrast, applying the additional 12.3 N•m of isometric hip extension moment produced by the combination of gluteus maximus and semimembranosus reduced the arm forces by 22 percent of body weight.

DISCUSSION

The simulation results with the use of the static sagittal plane model of human standing predicted that the active hip extension moment and body posture would both significantly impact the support forces required from the arms to maintain a static upright standing posture. Increasing the active hip extension moment decreased the arm forces required to remain erect. Decreasing hip flexion angle (i.e., standing more erect) also decreased the required upper-limb support forces in the upright position. The experimental results are consistent with the relationships identified in the simulations. Data from the two case studies suggest that standing performance depends on both the extension moment generated by the hip muscles and the posture assumed. For any

Table 2.

Measured total upper-limb support forces with different hip extensor muscle combinations while standing as erectly as possible. “No stim” indicates that no hip extensor muscles were activated on the tested side. “All” indicates that all three available hip extensor muscles were activated. Small standard deviations relative to means indicate repeatability of responses and upper-limb support forces consistently decrease with increasing hip extension moment (number and strength of muscles activated).

Subject	Side		Total support force on arms (N) No stim	Glut+Add	Glut+Sem	Add&Sem	All
A	Right	μ	487	414	346	345	343
(12° hip flexion)		σ	(6)	(13)	(11)	(14)	(13)
	Left	μ	231	211	223	234	343
B		σ	(11)	(11)	(11)	(26)	(13)
(18° hip flexion)	Right	μ	401	364	224	—	—
		σ	(13)	(13)	(18)	—	—

μ=mean; σ=standard deviation; Glut=Gluteus Maximus; Add=Adductor Magnus; Sem=Semimembranosus.

given hip flexion angle between 5º to 25º, stimulation of the hip extensor muscles that produce the highest extension moment should yield standing with the least arm forces.

The effect of active hip extension moment on the upper-limb support forces depends on the hip flexion angle assumed while standing. The proportionality between the decreasing forces on the arms and the increasing hip extension moment derived from the stimulation results agrees with the experimental findings from subjects with SCI. The more erect the posture, the more sensitive arm support force is to hip extension moment and the more effective small increases in hip moment will be in producing significant changes in upper-limb support.

To decrease the forces imposed on supporting devices and to eventually free the neuroprosthesis user from reliance on their upper limbs for support during standing, one must maximize the moment generated by electrically activated hip extensors. Better activation may be achieved in the future with the use of nerve cuff electrodes or implantation of multiple electrodes in each hip extensor muscle. The former approach requires the development of new surgical approaches to the major nerves of the lower limbs, while the latter presupposes the existence of implantable stimulators with many more channels than are currently available.

Equally important for minimal upper-limb exertion, posture in terms of hip flexion angle, must be as erect as possible. By an individual achieving better posture, the moment needed to remain upright would decrease substantially. This result is unsurprising, but nonetheless highlights the need to maintain range of motion at the hip and take preventative measures to avoid hip flexion contractures, which are common after prolonged periods in the sitting position that follow long term SCI. It also reemphasizes the importance of avoiding inadvertent activation of the hip flexors during quiet standing. This commonly occurs via spillover to the rectus femoris or other structures innervated by the femoral nerve during stimulation of the quadriceps (sartorius), or other hip flexors innervated by lumbar nerve roots during stimulation of the trunk extensor muscles (psoas, iliacus, tensor fasciae latae). In addition to increasing the reliance on the upper limbs for support, FNS-induced standing with hip range of motion limitations or activation of the hip flexor musculature can lead to other deformities. For example, exaggerating the lumbar curve by pushing down on an assistive device to elevate the torso can compensate for active hip flexion or extension range limitations (anterior pelvic tilt), but the resulting lordodic posture is far from functional, and maintaining it can lead to long-term vertebral laxity or other back problems.

Future studies should address the role of the trunk in maintaining erect posture and minimizing reliance on support through the upper limbs. In particular, the effectiveness of stimulating the lumbar erector spinae muscles to stiffen the lower spine and rigidly couple the pelvis and torso needs to be investigated more thoroughly.

The static sagittal plane standing model has provided a first approximation of the forces that need to be imposed on the arms to keep the trunk erect as a function of applied hip extension moment. However, a dynamic model would be needed to account for movements and perturbations during standing and is essential for simulating the sit-to-stand transition. Although a planar approach is appropriate for the study of the sagittal components of support and posture, medio-lateral stability, abduction-adduction, and internal-external rotation are crucial for maintaining balance and should be addressed in future work with a more complex three-dimensional model. The static sagittal model used for these simulation studies also assumed no significant active moments or range of motion limitations at the ankle. Plantarflexion contractures are common after SCI and can significantly compromise upright standing posture by forcing subjects to hyperextend the knee or flex the hip to keep from falling backwards. Although the subjects in this study did not exhibit such ROM limitations, they should be taken into account in future studies in order to further generalize these results to a larger segment of the SCI population and explore the interaction of ankle and hip mechanics during upright stance. The ability to generalize the results could also be improved significantly by repeating similar experiments on additional subjects due to the high degree of intersubject variability.

Another possible source of error is the reliance on model parameter values derived from able-bodied individuals to gain insight into the standing performance of persons with SCI (34). In spite of the significant changes that take place following long-term paralysis, the modeling and simulation results derived from the able-bodied model can still provide insight into standing with FNS until anthropometric data that better represent the SCI population become available.

The absence of lower-limb sensation during standing may also lead subjects to adopt an exaggerated posture so that the arms bear more weight than necessary to provide stability. The absence of proprioception can cause the subjects to overly rely on a support device. This strategy is typically adopted to avoid falls and to reduce perturbation effects, but it can affect the relationship between the hip extension moment and the reduction of arm forces as measured in the experimental portion of this study.

Muscle spasms and fatigue may also have affected the measured support forces. Muscle spasms were usually observed in hip flexor muscles when the hip and knee were fully extended. All force records were screened prior to analysis to eliminate the data contaminated by muscle spasms. The effects of fatigue in hip extensors were minimized by controlling the duty cycle between successive bursts of muscle stimulation. Small and randomized duty cycles allowed the muscles to recover between contractions, and thus muscle forces were well maintained throughout the experimental sessions. Small postural adjustments during standing trials may also have produced some of the variability observed in the experimental data. Since these motions occurred primarily during the first second after the onset of a particular standing pattern, data screening and adjusting the averaging window to include only steady-state effects reduced their influence on the measurements.

This study suggests that adequate active hip extension strength is critical to FNS-induced standing performance. Lack of stimulated hip extension moment can be due to (1) reduced muscle strength due to deconditioning or atrophy after prolonged paralysis, (2) inability to recruit all the individual hip extensor muscles, and (3) partial activation of a muscle due to electrode design or placement. The results indicate that particular attention should thus be focused on methods for enhancing hip extension moments. Future studies should investigate methods to improve the recruitment properties of the hip extensor muscles through innovative designs of stimulating electrodes. Nerve cuff electrodes may more fully activate the muscles and increase the hip extension moment possible with FNS, or alternatively be used to selectively activate the three vasti while avoiding the active hip flexion moment produced by rectus femoris during quadriceps stimulation. Alternatively, the design of multichannel implantable stimulators should be improved to increase the number of stimulus channel available for the hip extensor muscles or to provide methods to provide the waveforms required for selective activation of targeted fascicles. Finally, hip extension might be further maximized (and arm forces reduced) by exploring more creative surgical interventions such as distal hamstring transpositions or proximal rectus femoris releases to augment the actions of FNS.

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