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Posture shifting after spinal cord injury using functional neuromuscular stimulation—A computer simulation study

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ABSTRACT

The ability for individuals with spinal cord injury (SCI) to affect changes in standing posture with functional neuromuscular stimulation (FNS) was explored using an anatomically inspired musculoskeletal model of the trunk, pelvis and lower extremities (LE). The model tracked trajectories for anteriorly and laterally shifting movements away from erect stance. Forces were applied to both shoulders to represent upper extremity (UE) interaction with an assistive device (e.g., a walker). The muscle excitations required to execute shifting maneuvers with UE forces $<10\%$ body-weight (BW) were determined via dynamic optimization. Nine muscle sets were examined to maximize control of shifting posture. Inclusion of the Psoas and External Obliques bilaterally resulted in the least relative UE effort (0.119, mean UE effort $=45.3\ \mathrm{N} \equiv 5.4\%$ BW) for anterior shifting. For lateral shifting, the set including the Psoas and Latissimus Dorsi bilaterally yielded the best performance (0.025, mean UE effort $=27.8\ \mathrm{N} \equiv 3.3\%$ BW). However, adding the Psoas alone bilaterally competed favorably in overall best performance across both maneuvers. This study suggests suitable activation to specific muscles of the trunk and LE can enable individuals with SCI to alter their standing postures with minimal upper-body effort and subsequently increase reach and standing work volume.

1. Introduction

Quiet standing and simple stepping maneuvers can be restored to individuals paralyzed by spinal cord injury (SCI) through neuroprostheses employing functional neuromuscular stimulation (FNS). Existing FNS systems maintain a single erect standing posture by continuously activating the knee, hip and trunk extensors (Jaeger et al., 1989; Yarcony et al., 1990; Triolo et al., 1996). Current systems essentially lock users into a single upright posture with no means to alter position except by pulling or pushing against the continuously activated muscles with the upper extremities (Kobetic et al., 1999). Furthermore, existing FNS systems are capable of activating only a small number of carefully selected muscles, making advanced functions and finer movements difficult. The purpose of this study was to examine the feasibility of dynamically shifting standing posture with low upper extremity exertion and a minimal number of optimally selected muscles. The ability to dynamically shift posture by appropriately modulating stimulation would expand work volume to allow users to reach and manipulate objects or prepare for anticipated disturbances, thus affording greater control over the environment and reducing the potential for falls.

Simplifying assumptions in prior modeling studies examining standing balance include: actuating the system by joint moments (Hemami and Wyman, 1979; Kim et al., 2006; Matjacic et al., 2001), representing the body as a multi-joint single inverted pendulum (Soetanto et al., 2001; Gollee et al., 2004), or combining pelvis and trunk into a single segment (Mihelj and Munih, 2004). While adequate for theoretical investigations into disturbance response and single limb stance, these models were essentially static oversimplifications. Exploring FNS-generated movement in three dimensions is important because muscle actions are not confined to single planes. For instance, stimulating the tibialis anterior after SCI causes the body to both fall forward (a sagittal plane movement) and lean sideways (a coronal plane movement). Moreover, the closed chain defined by maintaining the feet on the ground effectively reduces the system degrees of freedom but couples their individual effects across all movement planes. Thus, stimulation of any muscle about the closed chain can result in complex motor behavior unaccounted for by planar models. Finally, it is necessary to represent the pelvis and trunk separately since important muscles attach to one without spanning the other. A single pelvis–trunk segment could require complex synergy patterns from many muscles to constrain the anatomy to fit the simplified model.

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Recent simulation studies with 3D musculoskeletal models have demonstrated the possibility of holding the body statically at discrete bipedal postures with FNS (Heilman et al., 2006; Gartman et al., 2008). These studies used static optimization to determine optimal muscle sets to maintain the body in different postures. However, the muscle forces calculated to keep the body in static equilibrium are not guaranteed to be sufficient to move the body dynamically from one posture to another.

In our study, a 3D model was used to explore the feasibility of producing dynamic movements with FNS in a typical individual with paralysis. The results will inform future clinical implementation of neuroprostheses designed to restore standing balance after SCI and assist surgeons and rehabilitationists in the selection of optimal muscles and stimulation patterns.

2. Methods

2.1. Musculoskeletal model

The musculoskeletal model (Fig. 1) was a doubly supported inverted pendulum consisting of 21 bone segments connected by 21 joints (Zhao et al., 1998). The segmental mass and inertia properties were calculated according to anthropometric tables (Winter, 1990) based on an average healthy male (weight 840.7 N, height 1.72 m). The model was actuated by 32 Hill-type muscle elements (Zajac, 1989) including bilateral Medial Gastrocnemius (MEDGAS), Tibialis Anterior (TIBANT), Vastus Lateralis and Medialis (VASTI), Semimembranosus (SEMIMEM), Adductor Magnus (AMAG), Gluteus Medius (GMED), Gluteus Maximus (GMAX), Psoas (PSOAS), Erector Spinae (ESPINAE), External Obliques (EXTOBL), Latissimus Dorsi (LDORSI) and Sartorius (SART). For clarity, muscles on the left side were identified with a prefix (i.e., LMEDGAS). Broad hip muscles (GMAX, GMED and AMAG) were each modeled as two separate elements, but treated as single muscles by requiring them to have the same excitation.

Characteristics of the muscles were modified to match known changes that occur as a consequence of paralysis. Maximum moments generated with FNS following SCI are approximately 50% of able-bodied values (Heilman et al., 2006). To include this effect, the maximum isometric forces of all muscles in the model were reduced by half. Properties of the passive structures across the joints were experimentally determined from individuals with SCI (Amankwah et al., 2004; Lambrecht et al., 2009). To further capture the effects of SCI, the effort exerted by the upper extremities (UE) on a support device was included in the formulation of the mechanical system (Nataraj et al., 2010). UE effort was defined in terms of all six components (F_x^L, F_y^L, F_z^L, F_x^R, F_y^R, F_z^R) of the resultant forces (F^L and F^R) exerted at the left and right shoulders by voluntarily interacting with a walker.

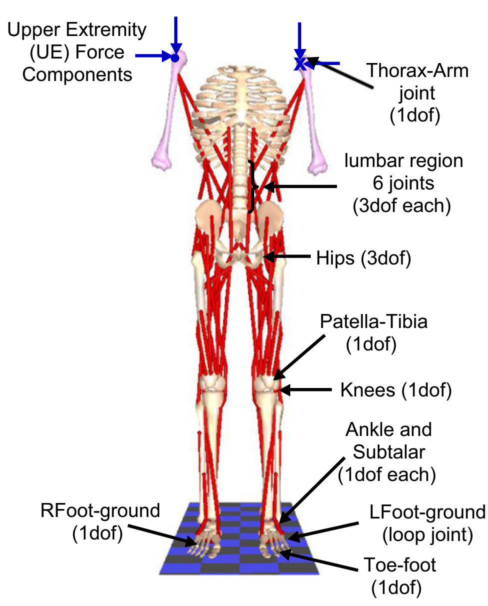

Fig. 1. Human bipedal musculoskeletal model used in the study showing the joints. Single degree-of-freedom revolute joints connected each foot to the talus to define the subtalar joints; the talus to the tibia/fibula to define the ankle joints, and the tibia/fibula to the femur to define the knee joints. Three degrees-of-freedom gimbal joints connected the two femurs to the pelvis to define flexion/extension, adduction/abduction and internal/external rotation of the hip joints. The torso attaches to the pelvis via a three degree-of-freedom gimbal joint that defined trunk flexion/extension (pitch), lateral bending (roll) and axial rotation (yaw). Five other lumbar joints also had three degrees-of-freedom each but moved synergistically in accordance with anatomically realistic kinematic constraints with respect to the torso–pelvic joint (White and Panjabi, 1990; Wilkenfeld et al., 2006). All other joints are prescribed to have zero velocity and acceleration; but were invaluable in defining realistic muscle wrapping points.

2.2. Muscle selection

Nine muscle sets (Table 1) built around a base group of 8 bilateral muscles (MEDGAS, TIBANT, VASTI, SEMIMEM, AMAG, GMED, GMAX and ESPINAE) were specified for the dynamic optimization. Muscles were added to the base set for specific functions: External Obliques for enhanced trunk flexion and lateral bending; Latissimus Dorsi for additional trunk extension; Sartorius for hip flexion and Psoas for hip and trunk flexion.

2.3. Dynamic optimization

Dynamic optimization can determine optimal muscle excitations and UE effort required to move the body from one position to another because the unknown control variables (muscle excitations and UE forces) are functions of time. This can be done by minimizing a scalar objective function subject to differential and non-differential (algebraic) constraints in the unknown variables.

2.3.1. Objective function

The scalar objective function for the optimization was defined to penalize excessive deviation from the desired trajectories throughout the path and any other function defined at the final time

$$J = [g(\vec{x}, \vec{u}, \vec{\pi})]_{t_f} + \int_{t_0}^{t_f} (\vec{x} - \vec{x}^d)^T Q (\vec{x} - \vec{x}^d) dt \tag{1}$$

Table 1
Muscle sets used in the optimization study. Each set consisted of the base set plus zero or more additional muscles chosen to enhance the capabilities of the base set.

SET	BASE SET	EXTOBL	LDORSI	SART	PSOAS
1	■				
2	■	■			
3	■		■		
4	■			■	
5	■				■
6	■	■		■	
7	■		■	■	
8	■	■			■
9	■		■		■

■ ACTIVE MUSCLE