Published in final edited form as:

Clin Biomech (Bristol, Avon). 2004 March ; 19(3): 313–321. doi:10.1016/j.clinbiomech.2003.12.005.

Femoral loads during passive, active, and active–resistive stance after spinal cord injury: a mathematical model

Laura A. Frey Law and Richard K. Shields*

Graduate Program in Physical Therapy and Rehabilitation Science, The University of Iowa, 1-252 Medical Education Building, Iowa city, IA 52242, USA

Abstract

Objective—The purpose of this study was to estimate the loading environment for the distal femur during a novel standing exercise paradigm for people with spinal cord injury.

Design—A mathematical model based on experimentally derived parameters.

Background—Musculoskeletal deterioration is common after spinal cord injury, often resulting in osteoporotic bone and increased risk of lower extremity fracture. Potential mechanical treatments have yet to be shown to be efficacious; however, no previous attempts have been made to quantify the lower extremity loading during passive, active, and active–resistive stance.

Methods—A static, 2-D model was developed to estimate the external forces; the activated quadriceps forces; and the overall bone compression and shear forces in the distal femur during passive (total support of frame), active (quadriceps activated minimally), and active–resistive (quadriceps activated against a resistance) stance.

Results—Passive, active, and active–resistive stance resulted in maximal distal femur compression estimates of ~45%, ~75%, and ~240% of body weight, respectively. Quadriceps force estimates peaked at 190% of body weight with active–resistive stance. The distal femur shear force estimates never exceeded 24% of body weight with any form of stance.

Conclusions—These results support our hypothesis that active–resistive stance induces the highest lower extremity loads of the three stance paradigms, while keeping shear to a minimum.

Relevance—This model allows clinicians to better understand the lower extremity forces resulting from passive, active, and active–resistive stance in individuals with spinal cord injury.

Keywords

Functional electrical stimulation (FES); Exercise; Bone; Osteoporosis; Paralysis; Rehabilitation; Compression; Shear

1. Introduction

Musculoskeletal deterioration is a common sequela of complete spinal cord injury (SCI) (Shields and Dudley-Javoroski, 2003). Skeletal muscle atrophies and transforms to fast fatigable fibers 12–18 months following injury (Shields, 1995, 2002). Similarly, lower extremity bone mineral density (BMD) rapidly decreases over the first 18 months after injury (Biering-Sorensen et al., 1990; de Bruin et al., 2000; Frey-Rindova et al., 2000;

Garland et al., 1992), resulting in significantly increased risk of lower extremity fracture (Frisbie, 1997).

Rehabilitation following SCI has historically centered on compensatory techniques to regain function; however, continued scientific advances are now shifting the foci of rehabilitation to include stressing the paralyzed limbs. The ultimate goal of many scientists, clinicians, and persons with SCI is to normalize impaired function through functional electrical stimulation (FES), assisted ambulation techniques, and ultimately repair of the spinal cord itself. Unfortunately, many people with SCI today may not have the muscle force-generating capacity or the skeletal integrity to benefit from these approaches if the musculoskeletal system is allowed to deteriorate.

Electromechanical interventions have been shown to induce muscle adaptations following SCI (Belanger et al., 2000; Sloan et al., 1994), partially preventing and/or reversing the atrophy. However, no loading interventions thus far have been shown to have more than minimal, if any, effects on lower limb osteoporosis after SCI (BeDell et al., 1996; Belanger et al., 2000; Biering-Sorensen et al., 1988; Bloomfield et al., 1996; Dauty et al., 2000). The actual loads from these studies were not quantified and may have been below an osteogenic threshold. In an effort to promote an osteogenic response, we recently developed an exercise paradigm that strives to increase the compressive loads on the bone while minimizing the shear forces.

According to Frost's mechanostat theory (Frost, 1987), there is an optimal range of mechanical loading essential for osteogenesis. Direct, in vivo measurements are difficult to obtain and may incur significant medical risk to subjects. Thus, we developed a mathematical model so that clinicians or researchers can better understand the lower extremity forces occurring during passive, active, and active-resistive stance in individuals with SCI. This knowledge is necessary to accurately establish if various doses of mechanical loading can offset the musculoskeletal deterioration that can occur after SCI.

Accordingly, the purpose of this study was to develop a mathematical model to estimate the loading environment for the distal femur during passive, active, and active-resistive stance for people with SCI. The specific aims were to (1) predict the external forces applied to the lower extremity during passive stance, (2) estimate the quadriceps force during active and active-resistive stance, and (3) estimate distal femur compression (axial) and shear (transverse) forces for passive, active, and active-resistive stance. We hypothesized that active-resistive stance would produce the greatest quadriceps muscle and bone compressive forces, while limiting the potentially deleterious shear forces.

2. Methods

2.1. Standing system

We developed an active and active-resistive standing system, which utilized a three-point support system common to many commercially available standing frames (Gear et al., 1999; Walter et al., 1999). A belt supports the hips posteriorly, pads support the knees anteriorly, and the feet are located on the floor. This system allows individuals with complete paralysis to stand passively and subsequently load their lower extremities (see Fig. 1 for a schematic representation of stance postures).

To perform active and active-resistive standing, a computer-controlled, electrical, muscle stimulator is used to activate the subjects' quadriceps muscles for additional lower extremity loading and/or muscle training purposes. The quadriceps muscles are stimulated with the use of surface, pad electrodes placed over the proximal and distal thighs, bilaterally. The

stimulator is programmed to deliver 20 Hz trains for 5 s followed by 5 s of rest. Typically, after numerous training sessions, subjects are able to tolerate three bouts of 60 contractions, with 5-min rest intervals, without difficulty.

When sufficient, this electrically induced knee extension moment replaces the support required from the kneepads in the passive condition, while the hip belt remains taut. To further resist this knee extension moment, a custom-built mechanical braking system and load cell applies a horizontal force to the back of the knee. Varying resistance levels may be used to maximize quadriceps muscle training at various stance positions under isometric conditions, resulting in active–resistive stance. The load cell measures the magnitude of the knee extension resultant force generated electrically via the quadriceps. Thus, passive standing requires total support of the standing frame. Active standing utilizes adequate quadriceps activation so that the anterior knee support is not necessary (no added resistance). Active–resistive stance requires quadriceps activation, in concert with resisted knee extension, to register a force reading on the load cell.

2.2. Mathematical model

A static, bilaterally symmetric, 2-D model was developed to estimate the external forces, the activated quadriceps forces and the overall bone compression and shear forces in the distal femur during passive, active, and active–resistive standing. The model consists of a three-bar linkage representing the thigh, shank, and foot segments of one lower limb (see Fig. 2 and nomenclature), utilizing anthropometric values for link lengths, masses, and center of mass locations derived from subject height and weight (Chaffin and Andersen, 1998). All results contained in this report are based on a hypothetical subject (height = 1.73 m, weight = 68.04 kg). One-half of the weight of the head, arms, and trunk acts as a single external force applied downward at the hip; the hip resultant moment is assumed negligible; the belt force acts at a slightly upward angle at the hip; the kneepad force acts horizontally at the knee (friction assumed negligible); and the ground reaction force components act on the lower surface of the foot segment. Six standing postures (i.e. combinations of belt, thigh segment, and shank segment angles) were modeled to estimate the range of forces experienced by a subject performing this exercise at varying degrees of upright stance (see Fig. 1).

The joints at the knee and the ankle were modeled as frictionless 2-D hinge joints with no internal moments in the passive condition, and only a knee internal moment in the active conditions, due solely to the electrically activated quadriceps muscle. The rectus femoris portion of the quadriceps crosses the hip joint, and although electrically activated, the internal hip joint moment was assumed negligible. Observations supported that the hip moment was minimal with quadriceps activation because the trunk remains stable. For purposes of balance, the upper extremities are allowed to occasionally touch a supporting surface, but these forces are intermittent and were considered negligible.

In other lower extremity models of gait and sit-to-stand activities, often the major difficulty is the determination of internal muscle forces. In subjects with complete SCI at the thoracic level or above, there are no voluntary muscle forces present in the lower extremities (including the hip musculature). All internal muscle forces are due to electrical stimulation, neuromuscular spasticity, or passive tension. Only electrically stimulated muscle forces were considered in this study. We chose to estimate quadriceps forces in the active state as a single vector, equal in magnitude to the supporting kneepad force during passive state plus any added resistance provided by the brake mechanism. This modeling approach necessitated that the forces in the quadriceps and patellar tendons were sufficient to maintain the upright posture (active stance) without an anterior knee supporting force (Fig. 2). The resultant of the quadriceps and the patellar tendon forces at the knee was assumed to act

horizontally, and exist only in the active and active–resistive conditions (all internal muscle and tendon forces were assumed negligible in the passive state). The angular positions of the three lower limb segments and the ground reaction forces were unchanged at the instant the passive exercise became “fully” active (i.e. no supporting kneepad force required to maintain standing). When the pad force fell to 0 the quadriceps produced an isometric contraction so that the angular segments were unchanged. Active stance was modeled with no additional knee resistance. Active–resistive stance was modeled with unilateral knee resistances of 16.7%, 33.3%, 50.0% and 67.7% of body weight (111.2, 222.5, 333.7, and 445.0 N for a 68.04 kg person). Uniform compression and shear of the distal femur was calculated and analyzed as the primary loading conditions of interest for these passive, active, and active–resistive standing exercises (see Fig. 3).

2.2.1. Passive condition equations—Using the constraint equations located in Appendix A, the three unknowns for the passive stance model, F_{pad} , F_{belt} , and N (F_x is modeled as a function of N , see nomenclature, are determined using the model’s three independent, scalar equilibrium equations. The sum of all external forces in the x -direction equal zero (Eq. (1)).

$$F_{\text{belt}} * \cos(\text{belt}\angle) - F_{\text{pad}} + \mu_{\text{gr}}N = 0 \quad (1)$$

The sum of all external forces in the y -direction equal zero (Eq. (2)).

$$N + F_{\text{belt}} * \sin(\text{belt}\angle) - 1/2W_{\text{hat}} - W_{\text{th}} - W_{\text{sh}} - W_{\text{ft}} = 0 \quad (2)$$

The sum of all external moments about the hip equal zero (Eq. (3)).

$$\begin{aligned} & -W_{\text{th}} * (L_{\text{th}} - CM_{\text{th}}) * \cos(\text{th}\angle) - F_{\text{pad}} * L_{\text{th}} \\ & * \sin(\text{th}\angle) - W_{\text{sh}} * \{L_{\text{th}} * \cos(\text{th}\angle) - CM_{\text{sh}} \\ & * \cos(\text{sh}\angle) + (N - W_{\text{ft}}) * \{L_{\text{th}} * \cos(\text{th}\angle) - L_{\text{sh}} \\ & * \cos(\text{sh}\angle) + d_x + F_x * \{L_{\text{th}} * \sin(\text{th}\angle) + L_{\text{sh}} \\ & * \sin(\text{sh}\angle) + L_{\text{ft}}\} = 0 \end{aligned} \quad (3)$$

Matlab computer software (Prentiss Hall, New Jersey, USA) was utilized to solve all model equations. The Matlab software solves these equations given the parameter inputs of segment angles, body height and weight, and the functional coefficient of friction (μ_{gr}) determined experimentally.

2.2.2. Active and active–resistive condition equations—Using the results from the passive model, the active model employs the following assumptions. First, the tension in the quadriceps tendon (F_{quad}) and the patellar tendon (F_{pat}) are not necessarily equal (Dahlkvist et al., 1982; Ellis et al., 1984) but their vertical components are assumed equal and opposite. If we assume that their combined internal angle is $\sim 10^\circ$ less than the internal knee angle (van Eijden et al., 1985, see Fig. 2) and we simply divide this 10° equally between the two tendon angles, then a geometric relationship between F_{quad} and F_{pat} follows (Eq. (4)).

$$F_{\text{pat}} = F_{\text{quad}} * \sin(\text{th}\angle - 5^\circ) / \sin(\text{sh}\angle - 5^\circ) \quad (4)$$

The horizontal resultant (F_{qres}) of F_{quad} and F_{pat} is shown in Eq. (5).

$$F_{\text{qres}} = F_{\text{quad}} * \cos(\text{th}\angle - 5^\circ) + F_{\text{pat}} * \cos(\text{sh}\angle - 5^\circ) \quad (5)$$

At the precise instant that the electrically induced quadriceps muscle force is sufficient to maintain stance without any external knee support, F_{qres} is assumed to be equal to the

passive condition pad force (active stance) plus any added horizontal resistance (active–resistive stance, see Eq. (6)). Thus, at this time the quadriceps develops isometric tension resulting in no change in segment angles, and the added resistance is instantly applied in the active–resistive condition.

$$F_{\text{qres}} = F_{\text{pad}} + R \quad (6)$$

Using Eqs. (4)–(6), Matlab solves for F_{quad} given F_{pad} , R , and the segment angles. This estimate for the electrically activated quadriceps force is determined without any direct measurements of the muscle, including cross-sectional area or electromyographic signals.

2.2.3. Compression and shear estimates—Using the forces modeled for the passive, active, and active–resistive stance conditions, uniform estimates for compression (F_c) and shear (F_v) in the distal femur were calculated using Eqs. (7) and (8) below (see Fig. 3). These loads were calculated at length $l = 0.85 * L_{\text{th}}$ (15% of femur length above knee) as this approximates a location on the distal femur where fractures commonly occur in individuals with SCI (Comarr et al., 1962). The femur is assumed to be uniform at this location.

$$\begin{aligned} -1/2 W_{\text{hat}} * \sin(\text{th}\angle) - F_{\text{belt}} * \cos(\text{th}\angle + \text{belt}\angle) \\ -0.85 W_{\text{th}} * \sin(\text{th}\angle) - F_{\text{quad}} * \cos(5) + F_c = 0 \end{aligned} \quad (7)$$

$$\begin{aligned} -1/2 W_{\text{hat}} * \cos(\text{th}\angle) + F_{\text{belt}} * \sin(\text{th}\angle + \text{belt}\angle) \\ -0.85 W_{\text{th}} * \cos(\text{th}\angle) + F_{\text{quad}} * \sin(5) - F_v = 0 \end{aligned} \quad (8)$$

2.2.4. Experimental measurements—Experimental measurements of the external forces were made to estimate realistic values for the model parameters (segment angles; functional coefficient of friction, u_{gr} ; and horizontal distance (d_x) between the normal force, N , and the ankle joint center of rotation (CoR)). Two subjects (one male with SCI, one able-bodied female) were asked to stand passively in a standing frame at a variety of possible stance postures (8 for the male subject, 6 for the female subject) after providing informed consent as approved by the institutional review board. Thigh and shank segment angles were determined using an Optotrak motion analysis system (Northern Digital, Waterloo, Ontario, Canada) in addition to being measured manually with a goniometer for each stance posture.

Normal ground reaction forces were measured using a Kistler forceplate (Kistler Inc, Amherst, New York, USA), which was calibrated prior to each subject's testing. The magnitudes of the normal and frictional forces (anterior/posterior) and the center of pressure location (CoP) were recorded for each standing posture tested. These values were used to estimate the functional coefficient of friction at each position ($F_x = u_{\text{gr}} * N$). The horizontal distance (d_x) from the force plate CoP to the ankle CoR, as determined by the Optotrak, was determined for each position tested.

Kneepad and belt force data were collected using uniaxial force transducers during passive standing. The belt force was measured using a Magnatek AWUI-50 load cell (Magnatek Inc., Chatsworth, California, USA) placed in series with the hip belt. The kneepad force transducer (Magnatek AWUI-250 load cell) was positioned unilaterally in line with a custom kneepad. Both the kneepad and belt force transducers were calibrated using multiple standard weight plates, using a standard ISA calibration testing protocol. A sensitivity analysis, using various d_x , u_{gr} , and belt force values, was completed.

3. Results

3.1. Experimental findings

The mean value for the functional coefficient of friction (u_{gr}) for both subjects over all 14 stance postures was 0.15, with a range of 0.04–0.29. The mean value for the distance from the foot CoP to the ankle CoR (d_x) for both subjects over all 14-stance postures was 3.6 cm, with a range of 2.8–4.4 cm. These mean experimental values were used in all subsequent model force predictions to simplify the model.

3.2. Model predictions

The external belt, kneepad and ground reaction forces are plotted vs. six stance postures in Fig. 4. The modeled belt, kneepad, and Normal ground reaction forces were relatively stable for the six stance postures, ranging from 19.5% to 23.9% of body weight, 26.9% to 28.7% body weight, and 41.3% to 49.5% body weight, respectively. Both the belt and kneepad forces decreased slightly, while the Normal ground reaction force increased with each increasingly upright posture. This is consistent with expectations that a more upright posture would need less support at the hips and knees than a more seated posture, and have more of the bodyweight transmitted through the lower limbs to the ground.

The estimates for the quadriceps muscle forces for the active ($R = 0$) and active resisted conditions ($R = 16.7\%$, 33.3% , 50% , and 67.7% of body weight) are shown in Fig. 5 vs. each of the six stance postures. The active muscle force was estimated to increase incrementally with each level of added resistance; the most upright postures required larger quadriceps forces for a given resistance level due to the decreasing mechanical advantage of the muscle (peaks at $\sim 190\%$ of body weight). That is, the horizontal resultant of the quadriceps and patellar tendons is a smaller percentage of the total muscle force as the stance postures become increasingly upright, thereby requiring greater total quadriceps muscle force to oppose a given resistance level with the most upright stance postures.

The compressive and shear forces at the distal femur were predicted for passive stance, active stance, and active–resistive stance (Fig. 6). The compression forces of the active condition ($R = 0$) were approximately double those predicted for the passive condition ($\sim 75\%$ vs. 45% of body weight, respectively). For active–resistive stance using an added resistance of 67.7% of body weight (445 N for a 68 kg individual), the compression forces reached $\sim 240\%$ of body weight or nearly six-times the passive condition estimate. The shear force estimates also increased with added resistance, but were so low during passive stance (range 1.1% to -7.9% body weight) that they only reached a maximum value of 24.0% body weight with $R = 67.7\%$ body weight, or one-tenth of the compressive force estimate.

3.3. Model sensitivity and validity

When the distance from the foot CoP to the ankle CoR (d_x) and the coefficient of friction (u_{gr}) were varied from -5.0 to 7.5 cm (in 2.5 cm increments) and 0.05 to 0.30 (in 0.05 increments), respectively, the average passive compression, passive shear, active compression, active shear and quadriceps force changed by less than 5% body weight per incremental parameter change for position 1. The sensitivity analysis for all six standing positions for d_x and u_{gr} is presented in Table 1.

When the active condition belt force was changed $\pm 15\%$ (in 5% increments) from the passive condition, the average shear and compressive forces changed from 0.25% to 0.31% body weight and 0.94% to 1.15% body weight, respectively, per incremental change in belt force. Thus, the shear force never exceeded 26.85% body weight when testing these variations in belt forces at a resistance level of 67.7% body weight.

The validity of this model cannot be tested non-invasively, however, we can estimate prediction errors by comparing the experimentally measured belt, pad and ground reaction forces with the modeled F_{belt} , F_{pad} and N forces. For the 14 stance postures tested using two subjects, the mean percent error (of absolute values) between the measured and modeled external forces were 17.4%, 17.8%, and 14.2% for the F_{pad} , F_{belt} , and N , respectively.

4. Discussion

The primary findings of this model are (1) external forces vary only minimally with stance position; (2) quadriceps forces are considerably influenced by stance position and resistance magnitude; and (3) compression in the distal femur is closely aligned with the electrically induced quadriceps muscle forces during active stance, approaching 240% of body weight, whereas shear never exceeds 25% of body weight.

We chose to model femoral compression and shear as the primary indicators of the loading environment due to the probable association between compressive loading and osteogenesis, as suggested by the mechanostat theory (Frost, 1987), as well as the potential deleterious effects of shear in individuals with SCI. Although compression and shear are certainly dependent upon local variations in anatomical structure and loading, these global estimates provide a novel attempt to evaluate the dose of a possible therapeutic intervention for individuals with SCI. Local strain estimates require extensive modeling of individual anatomical geometry and may not be as useful for categorizing the overall dose of an exercise intervention in a clinical trial. Bending moments were not modeled in this study, despite their known contributions to bone loading (Rubin et al., 1990) because they are particularly dependent on individual anatomical variations.

Passive, active, and active-resistive stance, with low levels of added resistance (16.7% body weight), induce femoral compression values less than 100% body weight for all six stance postures modeled. To achieve compression forces approaching loading levels commonly studied in able-bodied individuals (e.g. walking), active-resistive standing with a unilateral horizontal resistance of 33.3% body weight or greater was necessary. Indeed, Frost (1999) stated that muscles rather than body weight cause the greatest loads on bone. Additionally, basic animal studies have indicated that the mechanical threshold for osteogenesis may be as low as 130–300 microstrain and up to 1500–3000 microstrain depending on the frequency, strain rate, and duration of the stimulus (Frost, 1987). It is difficult to translate in vivo bone strains from animal work to a gross loading environment for humans. However, the pioneering work in animal models suggests that if the active-resistive standing exercise can indeed transmit loads at an appropriate frequency and strain-rate, compressive loads approaching 240% body weight may have the potential to be osteogenic.

FES cycling (Ragnarsson et al., 1988; Sloan et al., 1994) and quadriceps muscle training (Belanger et al., 2000) have been able to increase force-generating capability after SCI. Ragnarsson et al. (1988) also found muscular endurance to improve with training. Conversely, cycling with FES has been reported to induce only small improvements in BMD (Bloomfield et al., 1996; Hangartner et al., 1994; Mohr et al., 1997) as well as have no effect (BeDell et al., 1996; Bloomfield et al., 1996; Leeds et al., 1990; Sloan et al., 1994) on lower extremity BMD measurements in individuals with SCI. Additionally, neither passive standing, ambulation with long-leg braces, nor ambulation with FES have yet to exhibit any improvement in lower extremity BMD in chronically injured subjects (Biering-Sorensen et al., 1988; Dauty et al., 2000; Kunkel et al., 1993; Needham-Shropshire et al., 1997; Thoumie et al., 1995; Wilmet et al., 1995).

While it is possible that different stimuli are necessary to cause adaptations in muscle and bone after SCI, it is also possible that osteoporosis, unlike muscle atrophy, is essentially

irreversible once established. Prevention rather than treatment may have the greatest potential to alleviate this significant SCI complication. The subject populations of previous BMD studies were comprised almost exclusively of individuals with chronic rather than acute SCI (BeDell et al., 1996; Belanger et al., 2000; Bloomfield et al., 1996; de Bruin et al., 1999; Hangartner et al., 1994; Kunkel et al., 1993; Leeds et al., 1990; Mohr et al., 1997; Needham-Shropshire et al., 1997; Thoumie et al., 1995). Additionally, these interventional BMD studies may have utilized sub-threshold mechanical stimuli. The use of relatively low bone loading regimens is not unexpected due to the extensive atrophy of chronically paralyzed muscle (Shields, 1995, 2002) and concerns of fracture, which have been reported to occur with physical interventions (Hartkopp et al., 1998; Jones et al., 2002; Valayer-Chaleat et al., 1998).

The model results may not be representative of the entire population of persons with SCI. Standard anthropometric body segment values were based on uninjured subject populations, however, persons with spinal cord injury may have altered body composition (Jones et al., 1998; Spungen et al., 2000). Moreover, issues such as spasticity and contracture were not considered in this model. Two of the model parameter values (μ_{gr} and d_x) were derived experimentally from two subjects. While their results provide evidence that these model parameters have merit, it should not be assumed that they are representative of the population at large. However, as supported by our sensitivity analysis, the model results are robust among a range of d_x and μ_{gr} values.

This study provides novel estimates of quadriceps and bone loading forces resulting from a recently custom-designed passive, active, and active–resistive standing exercise system. This may be valuable information for the rehabilitation clinician and/or researcher to better understand the forces involved during active and passive standing. This model does not, however, predict the effects of stance on muscle or bone properties. Indeed, even the highest forces associated with active–resistive stance may not be capable of preventing osteoporosis after SCI. Investigations, within our laboratory, are underway to establish the loading dose necessary to prevent osteoporosis after spinal cord injury.

In addition to the potential benefits of lower extremity loading, standing has been reported to improve perception of health quality (Kunkel et al., 1993; Walter et al., 1999), decrease skin ulcers, and improve bowel and bladder programs (Walter et al., 1999). Thus, the standing methodology presented in this study may represent the ideal rehabilitative exercise program to prevent musculoskeletal deterioration following spinal cord injury.

5. Conclusion

This model provides a first attempt to estimate electrically induced quadriceps forces and compression and shear loads in the distal femur during a passive, active, and active–resistive stance exercise paradigm for individuals with SCI. Shear force predictions remain below 24% of body weight despite quadriceps and compression forces reaching 190% and 240% of body weight, respectively. These results support our hypothesis that active–resistive stance does induce the greatest lower extremity loads of the three stance exercise paradigms evaluated. Active–resistive stance may be relatively safe for individuals with SCI, given the low shear forces. Future studies will be necessary to determine if active or active–resistive stance have the potential to prevent musculoskeletal deterioration following SCI.

Acknowledgments

An award (NIH R01-HD39445) from the National Center for Medical Rehabilitation Research to Richard K. Shields supported this research. A doctoral training award from the Foundation for Physical Therapy supported Laura Frey Law during this study.

Appendix A. Constraint equations

The following constraint equations and conditions simplify the standing exercise so that the model equations can be solved explicitly.

1. $F_x = \mu_{gr}N$, where μ_{gr} is a functional coefficient of friction which is some unknown percentage of the actual coefficient of friction between a subject's foot and the floor, $\mu_{gr} = \beta\mu$, where $0 < \beta < 1$. A mean value, $\mu_{gr} = 0.15$, was determined experimentally.
2. The segment lengths, weights, and center of mass locations are assumed to be proportional to height (Ht) and weight (BW) using anthropometric standards (Chaffin and Andersen, 1998).
3. A mean horizontal distance from W_{ft} and N to ankle CoR ($d_x = 0.036$ m) was measured experimentally.
4. All external forces and segment angles are assumed to remain unchanged between passive and active conditions (except F_{pad} which is replaced by the internal quadriceps muscle force), as the transition is assumed to occur instantaneously.

References

- BeDell KK, Scremin AME, Perell KL, Kunkel CF. Effects of functional electrical stimulation-induced lower extremity cycling on bone density of spinal cord-injured patients. *American Journal of Physical Medicine and Rehabilitation*. 1996; 75:29–34. [PubMed: 8645435]
- Belanger M, Stein RB, Wheeler GD, Gordon T, Leduc B. Electrical stimulation: can it increase muscle strength and reverse osteopenia in spinal cord injured individuals? *Archives of Physical Medicine and Rehabilitation*. 2000; 81:1090–1098. [PubMed: 10943761]
- Biering-Sorensen F, Bohr H, Schaadt O. Bone mineral content of the lumbar spine and lower extremities years after spinal cord lesion. *Paraplegia*. 1988; 26:293–301. [PubMed: 3205570]
- Biering-Sorensen F, Bohr HH, Schaadt OP. Longitudinal study of bone mineral content in the lumbar spine, the forearm and the lower extremities after spinal cord injury. *European Journal of Clinical Investigation*. 1990; 20:330–335. [PubMed: 2114994]
- Bloomfield SA, Mysiw WJ, Jackson RD. Bone mass and endocrine adaptations to training in spinal cord injured individuals. *Bone*. 1996; 19:61–68. [PubMed: 8830990]
- Chaffin, D.; Andersen, G. *Occupational Biomechanics*. Williams & Wilkins; Baltimore: 1998.
- Comarr AE, Hutchinson RH, Bors E. Extremity fractures of patients with spinal cord injuries. *American Journal of Surgery*. 1962; 103:732–739. [PubMed: 13880733]
- Dahlkvist NJ, Mayo P, Seedhom BB. Forces during squatting and rising from a deep squat. *Engineering in Medicine*. 1982; 11:69–76. [PubMed: 7201419]
- Dauty M, Perrouin Verbe B, Maugars Y, Dubois C, Mathe JF. Supralesional and sublesional bone mineral density in spinal cord-injured patients. *Bone*. 2000; 27:305–309. [PubMed: 10913927]
- de Bruin ED, Frey-Rindova P, Herzog RE, Dietz V, Dambacher MA, Stussi E. Changes of tibia bone properties after spinal cord injury: effects of early intervention. *Archives of Physical Medicine and Rehabilitation*. 1999; 80:214–220. [PubMed: 10025500]
- de Bruin ED, Dietz V, Dambacher MA, Stussi E. Longitudinal changes in bone in men with spinal cord injury. *Clinical Rehabilitation*. 2000; 14:145–152. [PubMed: 10763791]
- Ellis MI, Seedhom BB, Wright V. Forces in the knee joint whilst rising from a seated position. *Journal of Biomedical Engineering*. 1984; 6:113–120. [PubMed: 6708484]
- Frey-Rindova P, de Bruin ED, Stussi E, Dambacher MA, Dietz V. Bone mineral density in upper and lower extremities during 12 months after spinal cord injury measured by peripheral quantitative computed tomography. *Spinal Cord*. 2000; 38:26–32. [PubMed: 10762194]
- Frisbie JH. Fractures after myelopathy: the risk quantified. *Journal of Spinal Cord Medicine*. 1997; 20:66–69. [PubMed: 9097259]

- Frost HM. Bone “mass” and the “mechanostat”: a proposal. *Anatomical Record*. 1987; 219:1–9. [PubMed: 3688455]
- Frost HM. Changing views about ‘Osteoporoses’ (a 1998 overview). *Osteoporosis International*. 1999; 10:345–352. [PubMed: 10591831]
- Garland DE, Stewart CA, Adkins RH, Hu SS, Rosen C, Liotta FJ, Weinstein DA. Osteoporosis after spinal cord injury. *Journal of Orthopaedic Research*. 1992; 10:371–378. [PubMed: 1569500]
- Gear AJ, Suber F, Neal JG, Nguyen WD, Edlich RF. New assistive technology for passive standing. *Journal of Burn Care and Rehabilitation*. 1999; 20:164–169. [PubMed: 10188115]
- Hangartner TN, Rodgers MM, Glaser RM, Barre PS. Tibial bone density loss in spinal cord injured patients: effects of FES exercise. *Journal of Rehabilitation Research and Development*. 1994; 31:50–61. [PubMed: 8035360]
- Hartkopp A, Murphy RJ, Mohr T, Kjaer M, Biering-Sorensen F. Bone fracture during electrical stimulation of the quadriceps in a spinal cord injured subject. *Archives of Physical Medicine and Rehabilitation*. 1998; 79:1133–1136. [PubMed: 9749697]
- Jones LM, Goulding A, Gerrard DF. DEXA: a practical and accurate tool to demonstrate total and regional bone loss, lean tissue loss and fat mass gain in paraplegia. *Spinal Cord*. 1998; 36:637–640. [PubMed: 9773449]
- Jones LM, Legge M, Goulding A. Intensive exercise may preserve bone mass of the upper limbs in spinal cord injured males but does not retard demineralisation of the lower body. *Spinal Cord*. 2002; 40:230–235. [PubMed: 11987005]
- Kunkel CF, Scremin AM, Eisenberg B, Garcia JF, Roberts S, Martinez S. Effect of standing on spasticity, contracture, and osteoporosis in paralyzed males. *Archives of Physical Medicine and Rehabilitation*. 1993; 74:73–78. [PubMed: 8420525]
- Leeds EM, Klose KJ, Ganz W, Serafini A, Green BA. Bone mineral density after bicycle ergometry training. *Archives of Physical Medicine and Rehabilitation*. 1990; 71:207–209. [PubMed: 2317139]
- Mohr T, Podenphant J, Biering-Sorensen F, Galbo H, Thamsborg G, Kjaer M. Increased bone mineral density after prolonged electrically induced cycle training of paralyzed limbs in spinal cord injured man. *Calcified Tissue International*. 1997; 61:22–25. [PubMed: 9192506]
- Needham-Shropshire BM, Broton JG, Klose KJ, Lebwohl N, Guest RS, Jacobs PL. Evaluation of a training program for persons with SCI paraplegia using the Parastep 1 ambulation system: part 3. Lack of effect on bone mineral density. *Archives of Physical Medicine and Rehabilitation*. 1997; 78:799–803. [PubMed: 9344296]
- Ragnarsson KT, Pollack SO, Daniel WJ, Edgar R, Petrofsky J, Nash MS. Clinical evaluation of computerized functional electrical stimulation after spinal cord injury: a multicenter pilot study. *Archives of Physical Medicine and Rehabilitation*. 1988; 69:672–677. [PubMed: 3262335]
- Rubin CT, McLeod KJ, Bain SD. Functional strains and cortical bone adaptation: epigenetic assurance of skeletal integrity. *Journal of Biomechanics*. 1990; 23(Suppl. 1):43–54. [PubMed: 2081744]
- Shields RK. Fatigability, relaxation properties, and electromyographic responses of the human paralyzed soleus muscle. *Journal of Neurophysiology*. 1995; 73:2195–2206. [PubMed: 7666132]
- Shields RK. Muscular, skeletal, and neural adaptations following spinal cord injury. *Journal of Orthopaedic and Sports Physical Therapy*. 2002; 32:65–74. [PubMed: 11838582]
- Shields RK, Dudley-Javoroski S. Musculoskeletal deterioration and hemi-corpectomy after spinal cord injury. *Physical Therapy*. 2003; 83:263–275. [PubMed: 12620090]
- Sloan KE, Bremner LA, Byrne J, Day RE, Scull ER. Musculoskeletal effects of an electrical stimulation induced cycling programme in the spinal injured. *Paraplegia*. 1994; 32:407–415. [PubMed: 8090549]
- Spungen AM, Wang J, Pierson RN Jr, Bauman WA. Soft tissue body composition differences in monozygotic twins discordant for spinal cord injury. *Journal of Applied Physiology*. 2000; 88:1310–1315. [PubMed: 10749824]
- Thoumie P, Le Claire G, Beillot J, Dassonville J, Chevalier T, Perrouin-Verbe B, Bedoisseau M, Busnel M, Cormerais A, Courtillon A, et al. Restoration of functional gait in paraplegic patients with the RGO-II hybrid orthosis. A multicenter controlled study II: Physiological evaluation. *Paraplegia*. 1995; 33:654–659. [PubMed: 8584300]

- Valayer-Chaleat E, Calmels P, Giraux P, Fayolle-Minon I. Femoral fracture and iatrogenic hyperthyroidism in spinal cord injury. *Spinal Cord*. 1998; 36:593–595. [PubMed: 9713932]
- van Eijden TM, de Boer W, Weijs WA. The orientation of the distal part of the quadriceps femoris muscle as a function of the knee flexion–extension angle. *Journal of Biomechanics*. 1985; 18:803–809. [PubMed: 4066722]
- Walter JS, Sola PG, Sacks J, Lucero Y, Langbein E, Weaver F. Indications for a home standing program for individuals with spinal cord injury. *Journal of Spinal Cord Medicine*. 1999; 22:152–158. [PubMed: 10685379]
- Wilmet E, Ismail AA, Heilporn A, Welraeds D, Bergmann P. Longitudinal study of the bone mineral content and of soft tissue composition after spinal cord section. *Paraplegia*. 1995; 33:674–677. [PubMed: 8584304]

Nomenclature

W_{hat}	weight of head, arms and trunk
W_{th}	weight of thigh
W_{sh}	weight of shank
W_{ft}	weight of foot
belt ∠	angle of belt vs. horizontal
th ∠	thigh segment angle vs. horizontal
sh ∠	shank segment angle vs. horizontal
knee ∠	thigh∠+shank∠
BW	body weight
CoR	joint center of rotation
F_{c}	distal femur compressive force
M_{b}	femoral bending moment (not calculated)
F_{belt}	tension in hip support belt
F_{pad}	horizontal Kneepad force
N	normal ground reaction force
F_{x}	frictional force at ground (assumed = μN)
F_{quad}	force in quadriceps tendon
F_{pat}	force in patellar tendon
R	added resistance in x -direction
F_{gres}	resultant of F_{quad} and F_{pat}
d_{x}	distance from ankle CoR to N
μ_{gr}	coefficient of friction at ground
F_{v}	distal femur shear force

Position #	1	2	3	4	5	6
Thigh angle (ϕ)	55°	59°	63°	67°	71°	75°
Shank angle (α)	75°	77°	79°	81°	83°	85°
Belt angle (β)	20°	16°	12°	8°	4°	0°

Fig. 1.

Schematic representations of six possible stance postures for individuals with spinal cord injury (SCI) using an external frame (not shown). The arrow at the hip and the vertical line at the knee represent a hip support belt and a knee support pad, respectively. The feet are placed on the ground.

Fig. 2. Schematic free body diagram of the lower body during (A) passive stance and (B) active (no added resistance, $R = 0$) and/or active-resistive stance ($R = 16.7\text{--}67.7\%$ BW). The internal forces from the quadriceps and patellar tendons (F_{quad} and F_{pat} , respectively) in (B) replace the F_{pad} in (A).

Fig. 3. Schematic free body diagram of the femur, sectioned at $l = 0.85$ of the total femur length, corresponding to the most common location of distal femur fractures after spinal cord injury. Distal femur shear (F_v) and compression (F_c) were estimated by this model, but the bending moments (M_b) were not calculated.

Fig. 4.

The model estimates for the external belt (F_{belt}), kneepad (F_{pad}), and ground reaction forces (N and F_x) during passive stance at each of the six stance postures (as shown in Fig. 1) are plotted in terms of percent body weight (% BW).

Fig. 5. Modeled quadriceps forces are shown in percent body weight (% BW) during active and active-resistive stance with resistances ranging from 13.7% BW to 67.7% BW for each of the six stance postures.

Fig. 6. Modeled distal femur compression (A) and shear (B) in terms of percent body weight are plotted for passive, active, and active-resistive stance at each of the six stance postures. Note the different scales between graph A and B.

Table 1

The mean change in the compression, shear, and quadriceps force during passive (P) and active (A) stance in position 1 and 6 for a given change in d_x (2.5 cm) or u_{gr} (0.05)

	Mean difference per 2.5 cm change in d_x (range = -0.05 to 7.50 cm)	Mean difference per 0.05 cm change in u_{gr} (range = -0.05 to 0.30 cm)
P_{comp}	0.66–0.76% BW	0.61–0.77% BW
P_{shear}	2.47–2.84% BW	2.295–2.88% BW
A_{comp}	3.12–6.54% BW	4.49–11.44% BW
A_{shear}	2.68–3.35% BW	2.68–3.80% BW
F_{quad}	2.46–5.80% BW	4.49–10.72% BW

P_{comp} = passive compression; P_{shear} = passive shear; A_{comp} = active compression; A_{shear} = active shear; F_{quad} = quadriceps force.