A biomechanical analysis of exercise in standing, supine, and seated positions: Implications for individuals with spinal cord injury

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Context/objective: The distal femur is the primary fracture site in patients with osteoporosis after spinal cord injury (SCI).

Objective: To mathematically compare the compression and shear forces at the distal femur during quadriceps stimulation in the standing, supine, and seated positions. A force analysis across these positions may be a consideration for people with SCI during neuromuscular electrical stimulation of the quadriceps.

Design: A biomechanical model.

Setting: Research laboratory.

Outcome measures: Compression and shear forces from the standing, supine, and seated biomechanical models at the distal femur during constant loads generated by the quadriceps muscles.

Results: The standing model estimated the highest compressive force at 240% body weight and the lowest shear force of 24% body weight at the distal femur compared with the supine and seated models. The supine model yielded a compressive force of 191% body weight with a shear force of 62% body weight at the distal femur. The seated model yielded the lowest compressive force of 139% body weight and the highest shear force of 215% body weight.

Conclusions: When inducing a range of forces in the quadriceps muscles, the seated position yields the highest shear forces and lowest compressive forces when compared with the supine and standing positions. Standing with isometric contractions generates the highest compressive loads and lowest shear forces. Early active resistive standing may provide the most effective means to prevent bone loss after SCI.

Keywords: Biomechanics, Compressive strength, Electrical stimulation, Shear strength, Fracture, Exercise regimens, Electrical stimulation, Spinal cord injuries

Introduction

Spinal cord injury (SCI) affects 11 000 individuals a year in the United States¹ resulting in severe muscle atrophy^{2,3} and bone loss.^{4,5} Six weeks after SCI, skeletal muscles below the level of the lesion decrease in cross-sectional area⁶ and begin a shift toward a fast fiber phenotype.⁷ Individuals with SCI will lose 50–60% of their bone mineral density (BMD)^{4,8} because of deterioration of the trabecular epiphyses^{9,10} and a thinning of the cortical wall.¹¹ Owing to this rapid musculoskeletal deterioration, individuals with

SCI have a 2% chance of sustaining a lower extremity fracture, which is double the fracture risk of the general population.¹² Medical advances in spinal cord regeneration (stem cells) may result in a future cure for spinal cord injury. However, those with chronic SCI will only be able to take advantage of such a breakthrough if rehabilitation attenuates the severe musculoskeletal deterioration.

Although there are many exercise interventions aimed at preventing musculoskeletal deterioration following SCI, it is important to understand the underlying stresses placed on the skeletal system during these activities. Compressive forces, which run parallel to the long axis of the bone, optimally load the bone and lead to advantageous adaptations;¹³ however, shear forces, which are

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forces perpendicular to the long axis of the bone, can be dangerous and cause fractures especially in osteoporotic bone.¹⁴ One of the most common sites of fracture for individuals with SCI is the distal femur.^{15,16} Bone loss in this region is extensive¹⁷ and neuromuscular electrical stimulation (NMES) can generate large shear forces capable of fracture.¹⁸

The position of the limbs when stress is applied determines the magnitude of the final compression and shear forces.^{19,20} Accordingly, an intervention that minimizes shear and maximizes compression loads is likely most appropriate for those with SCI.

NMES is a suitable strategy to load the musculoskeletal system of individuals with SCI. NMES can cause an increase in muscle cross-sectional area,^{21,22} an increase in muscle oxidative capacity,^{23,24} and retention of the endurance of the muscle fiber.²⁵ The underlying skeletal system also responds to the loads placed on the bone via electrically induced muscle contractions.^{26,27}

We recently showed a 31% increase in BMD following unilateral NMES of the soleus muscle compared with the untrained limb of the same individual.^{28,29} Importantly, the dose of compressive load to the tibia was greater than one time body weight ($\sim 100\text{--}150\%$ BW).²⁸ Indeed, there is a 'knowledge gap' in the clinic regarding the dose of stress exerted onto the skeletal

system during regularly prescribed exercises for people with SCI.

In an effort to narrow this knowledge gap, we evaluated several biomechanical models of exercise positions reported for those with SCI. Accordingly, the purpose of this study was to model the internal compressive and shear forces at the distal femur for the standing, supine, and seated positions across a range of quadriceps muscle forces. The general hypothesis is that the standing model will show the highest compression loads and the lowest shear forces for a given muscle force, which may indicate that standing compared with supine or seated position is more optimal for exercise in people with SCI.

Methods

Biomechanical model

Two-dimensional, static, bilaterally symmetric, biomechanical models were used for the standing, supine, and seated positions. All three models were rigid bodies consisting of one lower extremity represented as a three-bar-linkage system composed of the thigh, shank, and foot segments.^{26,27} The thigh (ϕ) and shank angles (α) used in the standing model (Fig. 1A) were replicated in the supine model (Fig. 1B). For the

Figure 1 Schematic free body diagrams of the external forces and loading conditions acting on a lower extremity for (A) the standing model,³² (B) the supine model, and (C) the seated model. (D) Schematic free body diagram of the distal femur which has been cut at 85% of thigh segment length from the hip joint. This location is a common site of fracture for those with SCI. The bending moment at the distal femur was not determined but the compressive force (F_c) and shear force (F_v) were calculated.

seated model (Fig. 1C), the thigh and shank angles remained constant, $\varphi = 90^\circ$ and $\alpha = 0^\circ$.

Since the force transducer was located at the knee for the standing model but at the distal shank for the supine and seated models, it was necessary, for the purpose of this comparison, to assume the quadriceps muscle force was uniform across conditions. However, a range of muscle forces were modeled. The quadriceps tendon, F_{quad} , and patellar tendon, F_{pat} , were not modeled as a simple pulley but rather, the horizontal components of F_{quad} and F_{pat} are assumed equal and opposite^{30,31} similar to the standing model where the vertical components were assumed to be equivalent.³² Although this assumption is based on non-SCI subjects, the standing model was previously validated using a subject with SCI. In addition, others report no significant differences in the patellar tendon moment arm between SCI and non-SCI.³³ Subsequently, if the combined internal angle of the quadriceps tendon and the patellar tendon is assumed to be less than 10° ,³⁴ then its bisection results in equation (1). (All nomenclature and equations are contained in the Appendix.)

Anthropometric values based on height and weight were used to determine masses, lengths, and center of masses for all the segments³⁵ from a hypothetical subject with a height of 1.73 m and weight of 68.04 kg. The knee and ankle joints were modeled as 2D, frictionless, pin joints, and the hip joint was fixed in the x -direction, $F_{x\text{hip}}$, and the y -direction, $F_{y\text{hip}}$. Although rectus femoris crosses the hip, the internal moment at the hip was assumed to be negligible. When the quadriceps muscles were activated, they resulted in forces acting through the quadriceps tendon, F_{quad} , and patellar tendon, F_{pat} . The internal moment at the knee was due to the force from the quadriceps muscle, therefore, is already included in the equilibrium equations. The resistive force, R_A , that opposed the quadriceps was modeled at the distal shank and acted at a distance L_m away from the lateral femoral condyles.

Each model was governed by three independent, static equilibrium equations. These equations were solved using Matlab (Prentiss Hall, NJ, USA) to determine the three unknown external forces $F_{x\text{hip}}$, $F_{y\text{hip}}$ and R_A . The unknown forces were determined by solving equations (2)–(4) for the supine model and solving equations (5)–(7) for the seated model. The thigh segment was then cut at 85% of the thigh segment length measured from the hip which is the most common site of fracture in individuals with SCI¹⁵ (Fig. 1D). The internal forces of compression, F_c , and shear, F_v , were estimated at this location. Equations (8)–(11) were solved to determine the compressive and

shear forces for the supine and seated models, respectively. The compressive and shear forces were calculated for the standing model using equations previously published from our laboratory.³² We previously completed a sensitivity analysis for the standing model which explored quadriceps muscle, compression, and shear force changes in response to varying the distance between foot center of pressure to ankle center of rotation and the coefficient of friction. The forces in the standing model were validated using experimental testing. The maximum percent error remained below 18%.³²

Data reduction and statistical analysis

The mean compressive and shear force for each position was calculated across all quadriceps muscle forces. Paired t -tests were used to compare the compressive and shear forces between the positions. When data were not normally distributed (shear force) the Mann–Whitney rank sum test was used. A value of $P < 0.05$ was considered statistically significant.

Results

Compressive and shear force outcomes

A range of quadriceps forces were analyzed across the standing, supine, and seated positions. For the standing model, the compressive forces range from 75 to 240% body weight for a quadriceps force range 30–190% body weight, respectively. Across these same ranges of quadriceps force, the supine model had compressive forces from 30 to 190% body weight and compressive forces for the seated model ranged from 20 to 140% body weight. The compressive forces for the standing, supine, and seated models are plotted against quadriceps force in Fig. 2. The compressive force for the

Figure 2 The estimated compressive forces at the distal femur for the standing, supine, and seated models plotted in terms of body weight (%BW) with respect to quadriceps force.

Figure 3 The estimated shear forces at the distal femur for the standing, supine, and seated models plotted in terms of percent body weight (%BW) as a function of quadriceps force.

standing model was 50% greater than the supine model and 130% greater than the seated model.

The shear force was also estimated at the distal femur for each of the three positions (Fig. 3). For quadriceps forces between 30 and 190% body weight, the standing model had a shear force that ranged from 2 to 24% body weight, respectively. The supine model had shear forces of 17–62% across the same quadriceps force range. The peak shear forces ranged from 40 to 215% body weight for the seated position when quadriceps muscle forces ranged from 30 to 90% body weight. The supine model and the seated model showed a 310 and 950% greater shear force than the standing model, respectively.

Fig. 4 shows the mean compressive and shear forces for the standing, supine, and seated positions when aggregated across all quadriceps muscle forces. Compressive force decreases from 131% body weight in the standing position to 94 and 66% body weight for the supine and seated positions ($P < 0.001$), respectively. Shear force increases from 10.9% body weight in the standing position to 37 and 102% body weight for the supine position and seated positions, respectively ($P < 0.001$). The compressive force decreases from 94 to 66% body weight when changing from the supine to the seated conditions ($P < 0.005$). In addition, the shear force increases from 37 to 102% body weight when changing from the supine to the seated condition ($P < 0.001$).

An example of the trabecular structure is shown from a computed tomography (CT) image of the distal femur from an individual with long-standing SCI and an individual without SCI (Fig. 5A and B).

Figure 4 (A) Mean compressive force and (B) mean shear force for the standing, supine, and seated positions. *Significant difference $P < 0.001$; # significant difference $P < 0.005$.

Discussion

The primary findings of this study support the hypothesis that the standing model generates the largest compressive forces and the smallest shear forces when compared with seated and supine exercise. The distal femur site was chosen because it is a frequent site of fracture for people with SCI¹⁵ as illustrated by the trabecular deterioration in the CT image of the single subject (Fig. 5A and B). The forces measured by our model are important because compressive load is believed to be osteogenic,^{26,36} while shear forces have been associated with fracture after SCI.¹⁴ Using the only published report of a fracture during an exercise protocol,¹⁴ we now know that a high shear force likely contributed to the fracture in the seated position. The biomechanical analysis in this report provides, for the first time, a direct estimate of the stress during seated, supine, and standing exercise in people with SCI.

Closed versus open kinetic chain exercise

The standing and supine models estimated compressive loads that have recently been shown to prevent bone loss

Figure 5 Trabecular architecture rendered from a cylindrical volume of interest of CT images at the distal femur of those (A) without SCI and (B) with SCI.

in people with SCI.³⁶ The supine condition generated 62% body weight shear while the standing condition never exceeded 24% body weight. The supine model is an open-kinetic chain exercise, while the standing model is a closed-kinetic chain exercise. Open-kinetic chain exercises produce higher shear forces at the knee^{37–40} while closed-kinetic chain exercises result in more compressive forces and joint stability while limiting shear forces.^{38,41}

The seated model provided the least ability to develop compressive load (140% BW) and the highest shear force (215% BW). Interestingly, the only reported fracture during exercise occurred in this same seated position.¹⁴ Importantly, the quadriceps force that we used was conservative because length–tension relationships support that greater quadriceps force is generated in the seated compared with supine position⁴² and the highest shear force occurs at 90° of knee flexion during seated open-kinetic chain exercises.^{19,40,43}

Implications of the biomechanical model findings

The greater compressive force for the standing model compared with the supine and seated models can be explained, in part, by the weight of the upper body during stance. Clearly, the combination of gravity and muscle force promotes a greater compressive load, which has already been shown to be osteogenic.^{44–47} Muscle contraction only, in the absence of resistive gravitational load during extended bed rest⁴⁸ or space-flight,^{49,50} results in a significant decrease in bone density. Importantly, muscle that is allowed to shorten against little or no resistance, contributes minimally to the mechanical loads that are known to be osteogenic.⁵¹ Because gravity contributes to the load during stance, higher compressive loads may be reached at lower levels of quadriceps muscle force. These findings support that resisting stance during muscle stimulation may provide the most feasible and effective method to introduce safe mechanical loads into the skeletal system.

People with SCI lack the capacity to stand because of severe orthostatic hypotension, but systolic blood pressure is increased when electrical stimulation is introduced during passive standing.⁵² If standing is not a feasible intervention, the findings from this modeling study supports that NMES in the supine position could be an effective temporary alternative. However, these models assert that optimizing compression and minimizing shear forces will ultimately be achieved, most efficiently, by active resistive stance for people with SCI. As shown in the models, the seated position yields the highest shear forces and lowest compressive force of all the exercise positions. Based on these findings, NMES during isometric contractions in the seated position is not recommended for people with SCI.

Although potentially high shear forces can be generated during seated exercise in people with SCI, many still commonly use this position to assess muscle physiological properties of the quadriceps.^{53–56} Using alternative stimulation frequency parameters to limit deleteriously high shear force should be considered when assessing physiological properties of paralyzed muscle.⁵⁷

Important considerations for the biomechanical models

Biomechanical models provide a valuable source to estimate loads in human tissues without risking injury to a subject. Several assumptions, however, limit the utility of mathematical models and need to be interpreted carefully. For example, the biomechanical models presented in this study were designed to represent the SCI population. However, databases of anthropometric values used in the model were determined from a non-SCI cohort, making a one to one confirmation unrealistic. For this reason, all of the models presented in this report make the same assumptions regarding limb mass, segment COM, and muscle force. Because of the extensive variation across individuals with SCI, the models presented in this report do not account for contractures, spasticity, or anatomical variants in insertion angles of skeletal muscle. Taken together, the biomechanical values in this report are meaningful to show a ‘relative’ difference between exercise conditions, but may not adequately reflect the absolute values.

We previously validated the isometric forces generated during stance in humans with SCI.³² A completely instrumented standing system was developed to validate the standing biomechanical model.^{32,58} This system allowed for unilateral activation and comparison of the opposite limb as a control.^{32,58} However, we did

not validate the supine and seated positions because of the proposed risk for those with SCI, and the published case of fracture during this position.¹⁴ Therefore, the seated and supine mathematical data provide a guideline to assist clinicians in understanding relative doses of stress generated on limb segments in people with SCI, but is limited because these models were not validated in humans with SCI.

Conclusion

Exercise is a ‘form of medicine’ for individuals with SCI. When prescribing a dose of stress to people with SCI it is important to have estimates regarding those stresses. In this study, we performed a head-to-head biomechanical comparison to ascertain which exercises deliver the greatest compressive and shear forces. Isometric contractions during the standing position demonstrated the highest compressive forces and the lowest shear forces at the distal femur, the site most commonly fractured in people with SCI. Isometric contractions in the supine position showed moderate compressive and shear forces, while the seated position showed deleteriously high shear forces and low compressive forces. Since the standing model is a closed-kinetic chain exercise, body weight is axially loaded over the distal femur which increases the compressive force and therefore limits shear forces. Taken together, this comparison suggests that caution should be exercised when generating high quadriceps forces through NMES during seated exercise in people with SCI.

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Appendix

Nomenclature

W_{HAT}	weight of the head, arms, and trunk
W_{th}	weight of the thigh
W_{sh}	weight of the shank
W_{ft}	weight of the foot
L_{th}	length of the thigh
L_{sh}	length of the shank
L_{ft}	length of the foot
CM_{th}	center of mass of the thigh
CM_{sh}	center of mass of the shank
R	force at the knee (standing model)
R_A	force at the tibial restraint (supine/seated models)
L_m	moment arm of quadriceps muscle torque

φ	angle from the thigh to the vertical
α	angle from the shank to the vertical
β	angle from tibial restraint to the shank
F_{xhip}	horizontal force at the hip
F_{yhip}	vertical force at the hip
F_{quad}	force in the quadriceps tendon
F_{pat}	force in the patellar tendon
F_c	compressive force at the distal femur
F_v	shear force at the distal femur

Equations

- $F_{quad}\sin(\varphi - 5) = F_{pat} \sin(\alpha - 5)$
- $$\sum F_x = 0; \quad F_{xhip} - F_{quad}\sin(\varphi - 5) + F_{pat} \sin(\alpha - 5) + R_A \sin \beta = 0$$
- $$\sum F_y = 0; \quad F_{yhip} - W_{th} - F_{quad}\cos(\varphi - 5) - F_{pat} \cos(\alpha - 5) - W_{sh} + R_A \cos \beta - W_{ft} = 0$$
- $$\sum M = 0; \quad -W_{th}(L_{th} - CM_{th})\sin \varphi - (F_{quad}\cos(\varphi - 5) + F_{pat} \cos(\alpha - 5))(L_{th} \sin \varphi) + (F_{quad} \sin(\varphi - 5) - F_{pat} \sin(\alpha - 5))(L_{th} \cos \varphi) - W_{sh}(L_{th} \sin \varphi + CM_{sh} \sin \alpha) + R_A \cos \beta(L_{th} \sin \varphi + L_m \sin \alpha) - R_A \sin \beta(L_{th} \cos \varphi + L_m \cos \alpha) - W_{ft}(L_{th} \sin \varphi + L_{sh} \sin \alpha + L_{ft}) = 0$$
- $$\sum F_x = 0; \quad F_{xhip} - F_{quad}\cos(5) + F_{pat} \sin(5) - R_A = 0$$
- $$\sum F_y = 0; \quad F_{yhip} - W_{th} - F_{quad} \sin(5) - F_{pat} \cos(5) - W_{sh} - W_{ft} = 0$$
- $$\sum M = 0; \quad W_{th}(L_{th} - CM_{th}) + (F_{quad} \sin(5) + F_{pat} \cos(5) + W_{sh} + W_{ft})L_{th} + R_A L_m = 0$$
- $$-F_{xhip} \sin \varphi - F_{yhip} \cos \varphi - F_{quad} \cos(5) + 0.85 W_{th} \cos \varphi + F_c = 0$$
- $$F_{xhip} \cos \varphi - F_{yhip} \sin \varphi - F_{quad} \sin(5) + 0.85 W_{th} \sin \varphi + F_v = 0$$
- $$-F_{xhip} - F_{quad} \cos(5) + F_c = 0$$
- $$-F_{yhip} + 0.85 W_{th} - F_{quad} \sin(5) + F_v = 0$$

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