INTRODUCTION

Spinal cord injury (SCI) injury is characterized by a rapid loss of bone mineral at sublesional regions. The clinical consequence of this bone loss is an increased lifetime risk for low-energy fracture that is substantially greater than the general population [1]. These fractures frequently occur around regions of the knee, e.g. the proximal tibia. Fractures after SCI are a source of considerable morbidity, loss of independence, and increased medical costs.

Although the time course and magnitude of bone loss following SCI has been well documented, the biomechanical relevance remains unclear. Bone fractures are ultimately biomechanical events, and small changes in bone mineral may have large mechanical consequences. Though not the standard of care, pharmaceutical treatment represents a potential therapeutic intervention to ameliorate bone loss and reduce fracture occurrence after SCI. However, SCI is also associated with structural changes in geometry and mineral distribution, and therefore pharmaceutical treatments that reverse bone loss alone may not necessarily restore mechanical integrity back to baseline levels.

Our purpose was to quantify changes in torsional stiffness and strength of the proximal tibia due to 1) actual bone loss and 2) simulated bone recovery in acute SCI.

METHODS

Ten adults with acute SCI consented to participate in this institutionally approved study. Subjects received two computed tomography (CT) scans of the knee (120 kV, 200 mAs, pixel resolution 0.352 mm, slice thickness 1 mm). Baseline scans were performed approximately 2 months after SCI and follow-up scans were performed approximately 4 months after baseline. All scans included a calibration phantom to convert CT Hounsfield units to bone apparent density $\rho_{\text{app}}$.

Physical and simulated models:

Proximal tibiae were segmented from the CT scans and four voxel-based geometric models were generated for each subject – two physical (baseline and follow-up) and two simulated (treatment 1 and treatment 2) models. The physical models were generated directly from baseline and follow-up scans. The simulated models represented hypothetical treatments that restored bone mineral back to baseline levels following the acute period of SCI. For treatment 1 models, the $\rho_{\text{app}}$ of voxels of follow-up models were uniformly increased so that bone mineral content (BMC) was equal to baseline levels. For treatment 2 models, the $\rho_{\text{app}}$ of voxels of follow-up models were also increased so that BMC was equal to baseline, however, the amount of mineral apposition was dependent on the available surface area for remodeling, or surface area density $S_v$. Martin [2] illustrated that bone $S_v$ (mm$^2$/mm$^3$) is related to $\rho_{\text{app}}$ (g/cm$^3$) by a 5th order polynomial:

$$S_v = 0.2 + 6.8\rho_{\text{app}} - 2.5\rho_{\text{app}}^2 + 2.3\rho_{\text{app}}^3 + 2.7\rho_{\text{app}}^4 - 0.9\rho_{\text{app}}^5.$$

Prediction of torsional stiffness and strength:

Torsional stiffness $K$ and strength $T_{\text{ult}}$ were predicted for each model using validated subject-specific finite element modeling procedures. Using cadaveric experimentation, these modeling procedures were able to predict in vitro torsional stiffness and strength with an $r^2$ of 0.95 and 0.91, respectively [3]. Briefly, the voxel-based models were converted to 8-node hexahedral elements with 1.5 mm edge lengths. Elements were assigned inhomogeneous nonlinear orthotropic material properties. Pre-yield elastic moduli in the axial direction $E_3$ (MPa) was defined as: $E_3 = 6570\rho_{\text{app}}^{1.37}$ [4]. Anisotropy was assumed the same throughout with $E_1 = 0.574E_3$, $E_2 = 0.577E_3$, $G_{12} = 0.195E_3$, $G_{23} = 0.265E_3$, $G_{31} = 0.216E_3$, $\nu_{12} = 0.427$, $\nu_{23} = 0.234$, and $\nu_{31} = 0.405$. The non-linear phase was modeled as bilinear elastic-plastic with a post-yield modulus 5% of the pre-yield modulus. Yield was defined by
Hill’s quadratic criterion for orthotropic materials. Yield strains were assumed isotropic in the normal (0.675%) and shear (1.215%) directions [5]. Models were loaded in torsion and K was quantified from the linear portion of the torque-rotation curve. The T_{ult} corresponded to the torque at which 10% of surface elements had failed, defined as a maximum principal strain greater than 1.410% [5].

**Data Analysis:**
Volumetric bone mineral density (vBMD) and BMC were calculated for each model. Trabecular (Tb) and cortical (Ct) specific BMD and BMC were also determined assuming a threshold of $\rho_{app}=1.0$ g/cm$^3$. Paired t-tests were used to examine differences in bone mineral and mechanical behavior between baseline and follow-up models as well as baseline and treatment models ($\alpha=0.05$).

**RESULTS AND DISCUSSION**
During the acute period of SCI, subjects lost a mean 8.3% of their vBMD and 8.9% of their BMC (Table 1). Although relative reductions in K were similar in magnitude to relative reductions in bone mineral, relative reductions in T_{ult} were some 1.5 to 2 times greater than the observed reductions in bone mineral (Fig 1). The discrepancy between relative reductions in K and T_{ult} is difficult to interpret, but it does indicate that bone loss had a larger influence on post-yield rather than pre-yield behavior.

Simulations of hypothetical treatments that restored bone mineral back to baseline levels suggested that the declines in proximal tibia mechanical behavior after acute SCI are potentially reversible; for these models K and T_{ult} were not significantly different from baseline. However, substantial scatter in the mechanical behavior after treatment was observed between subjects, and scatter-plots suggested that subjects who lost more than approximately 10% of their bone mineral by follow-up did not restore mechanical behavior back to baseline levels when bone mineral was recovered. Additionally, it should be noted that our modeling procedures cannot account for potential micro-structural changes in trabecular architecture, collagen cross-linking, and remodeling space.

The two distinct hypothetical treatments used to restore bone mineral back to baseline levels provides some insight into the mechanisms by which bone mineral is resorbed after SCI. In contrast to treatment 1 simulations, trabecular and cortical BMD and BMC for treatment 2 simulations were not significantly different from baseline. These data, therefore, suggest that the bone remodeling response after SCI is dependent to some degree on the available surface area for remodeling. Indeed, peripheral QCT studies of disuse have observed the greatest declines in bone mineral at regions expressing the largest surface area for remodeling [6].

**CONCLUSIONS**
Individuals with acute SCI experienced a marked loss of bone mineral at the proximal tibia. Losses in bone mineral were associated with a 1.5 to 2 fold reduction in torsional strength. Reductions in torsional strength may be reversible if pharmaceutical treatment is initiated soon after SCI.

**REFERENCES**

**ACKNOWLEDGEMENTS**
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**Table 1.** Percent change in bone mineral and mechanical behavior relative to baseline line (* p<0.05).  
<table>
<thead>
<tr>
<th></th>
<th>vBMD</th>
<th>BMC</th>
<th>Tb.vBMD</th>
<th>Tb.BMC</th>
<th>Ct.vBMD</th>
<th>Ct.BMC</th>
<th>K</th>
<th>T_{ult}</th>
</tr>
</thead>
<tbody>
<tr>
<td>Follow-up model</td>
<td>-8.3 (4.9)*</td>
<td>-8.9 (5.0)*</td>
<td>-10.3 (9.2)*</td>
<td>-9.7 (9.2)*</td>
<td>-1.1 (2.2)*</td>
<td>-8.0 (5.3)*</td>
<td>-9.9 (6.5)*</td>
<td>-15.8 (13.8)*</td>
</tr>
<tr>
<td>Treatment 1 model</td>
<td>0.7 (0.3)*</td>
<td>0.0 (0.0)</td>
<td>-6.0 (7.8)*</td>
<td>-7.2 (8.2)*</td>
<td>5.0 (4.8)*</td>
<td>7.0 (7.3)*</td>
<td>-2.2 (4.7)</td>
<td>-5.0 (10.0)</td>
</tr>
<tr>
<td>Treatment 2 model</td>
<td>0.7 (0.3)*</td>
<td>0.0 (0.0)</td>
<td>0.6 (4.8)</td>
<td>-2.9 (8.6)</td>
<td>0.6 (2.5)</td>
<td>0.4 (4.6)</td>
<td>-0.2 (3.4)</td>
<td>-2.4 (4.7)</td>
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**Figure 1.** Changes in T_{ult} were 2 fold greater than changes in vBMD. Insert of a finite element model with contour plot of max principal strain at T_{ult}.