Quantifying Stiffness During Downhill Running

J.J Chu, J. Hamill, G.E. Caldwell
Biomechanics Laboratory, Department of Exercise Science
University of Massachusetts, Amherst/USA

Introduction
Impact shocks have been shown to cause changes in joint characteristics similar to osteoarthritis, such as cartilage degeneration, subchondral bone stiffening, and trabecular microfracturing (Radin et al., 1973). The resulting shock wave initiated by heel strike can be altered as it travels upward through the body by both passive and active attenuation mechanisms. For example, mass-spring (MS) models and stride length/frequency manipulations have demonstrated that greater knee flexion at heel strike can reduce leg stiffness and increase shock attenuation (McMahon et al., 1987; Lafortune et al., 1996; Derrick et al., 2000). However, typically large kinematic changes are seen when stride length and frequency are manipulated. In contrast, downhill running at varying grades elicits a large range of tibial impact accelerations without inducing such large kinematic differences (Chu et al., 1999). Therefore it may be that stiffness modifications with changing grade during downhill running are different than with stride length alterations. Thus the purpose of this study was to investigate stiffness changes with increasing impact shock in a downhill running protocol.

Methods
Ten experienced runners ran on a treadmill at a fixed velocity (4.17 m/s) at 5 grades ranging from level to 12% downhill. Each subject was acclimatized to testing conditions and data were collected for 20 s at each grade in random order. Light weight accelerometers on the frontal bone of the head and distal anteromedial aspect of the tibia were tightened to subject tolerance to measure head (HA) and tibial (TA) accelerations (1000 Hz). Sagittal kinematics of joint markers were collected at 200 Hz to determine segmental COM parameters and initial conditions needed to model TA, HA and stiffness. A mass-spring-damper model (MSD) was developed to simulate both TA and HA during stance (Chu et al., 2000). A lower MSD complex (Figure 1) represented the support limb effective mass (M2), stiffness (K2) and damping (β). An upper MS represented the remaining body mass (M1) and stiffness (K1). K1, K2 and initial spring position (P2) were optimized using a Downhill Simplex algorithm and a cost function that gave the best fit between M2 acceleration and measured TA throughout stance. Summing the acceleration of M1 and M2 with an optimized shock transmission coefficient generated simulated HA. Exemplar data for both TA and HA are shown in Figure 2.

![Figure 1: MSD Model](image)

![Figure 2: Exemplar model (dark line) and experimental data(light line) for HA and TA](image)
Results & Discussion

Greater peak TA (23% higher than level, \( p = 0.02 \)) and HA (50% higher, \( p < 0.01 \)) were measured with steeper downhill grades (Table 1). However, the time to peak for both TA and HA did not change (\( p = 0.60 \) and \( p > 0.99 \)). The relatively larger change in HA compared to TA indicated less shock attenuation at the steeper grades (\( p < 0.01 \)).

The MSD model reproduced TA and HA waveforms well. Comparison of experimental and model data showed no statistical differences across grade conditions in impact peak (\( p = 0.12 \)), time of impact peak occurrence (\( p = 0.33 \)) or stance time (\( p = 0.22 \)). \( K_1 \) stiffness increased as downhill grade became steeper (\( p = 0.06 \)), but \( K_1 \) stiffness did not (\( p = 0.63 \)). A longer \( P_2 \) was generally seen with the greatest length occurring at the 9% grade (\( p < 0.02 \)). Greater model shock transmission coefficients were observed with steeper grades indicating less shock attenuation (\( p = 0.04 \)), in agreement with the experimental results.

The consistent \( K_1 \) was not surprising due to its relation with stance time which also remained constant across grades (\( p = 0.59 \)). Stepwise regression analysis showed that peak TA was determined by the interaction of \( K_1, K_2 \) and \( P_2 \) (\( r = 0.81; p < 0.01 \)) while the timing of peak TA was only influenced by \( K_2 \) (\( r = 0.46; p < 0.01 \)).

Derrick et al. (2000) reported that longer stride lengths resulted in an increase in both peak TA and stance time. This increase in peak TA was associated with a stiffer \( K_2 \) and the longer stance times with a more compliant \( K_1 \). In the present downhill running study, higher peak TA values were also accompanied by a stiffer \( K_2 \), but subjects maintained more consistent stance times and \( K_1 \) stiffness values. Therefore the downhill running protocol elicited different stiffness alterations than does stride length manipulation.

The model results suggest that peak TA and its timing were highly influenced by alteration of the support limb parameters (\( K_2 \) and \( P_2 \)) as impact shock increased. Thus if one could reduce \( K_2 \) stiffness peak TA should decrease, which may reduce the deleterious effects of impacts. However, as \( K_1 \) and \( K_2 \) were not associated with changes in shock attenuation and transmission, reducing \( K_2 \) stiffness may have no effect on shock attenuation.

References


<table>
<thead>
<tr>
<th>Grade</th>
<th>( K_1 ) (kN/m)</th>
<th>( K_2 ) (kN/m)</th>
<th>( P_2 ) (mm)</th>
<th>Shock Trans(%)</th>
<th>TA (g)</th>
<th>HA (g)</th>
<th>Stance (ms)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0%</td>
<td>76.90</td>
<td>108.14</td>
<td>8.71</td>
<td>23.89</td>
<td>7.86</td>
<td>1.77</td>
<td>188.00</td>
</tr>
<tr>
<td>3%</td>
<td>69.83</td>
<td>114.78</td>
<td>10.27</td>
<td>24.90</td>
<td>8.46</td>
<td>1.99</td>
<td>189.00</td>
</tr>
<tr>
<td>6%</td>
<td>71.46</td>
<td>106.10</td>
<td>11.43</td>
<td>28.34</td>
<td>9.26</td>
<td>2.45</td>
<td>192.00</td>
</tr>
<tr>
<td>9%</td>
<td>72.87</td>
<td>112.64</td>
<td>11.54</td>
<td>30.84</td>
<td>9.31</td>
<td>2.65</td>
<td>191.00</td>
</tr>
<tr>
<td>12%</td>
<td>73.73</td>
<td>126.62</td>
<td>10.49</td>
<td>29.05</td>
<td>9.64</td>
<td>2.62</td>
<td>191.00</td>
</tr>
</tbody>
</table>

Table 1: MSD model results with increasing downhill grade. \( K_1 \) and \( K_2 \) are the stiffness measures for the COM and support limb. \( P_2 \) is the initial spring position of \( K_2 \) and Shock Trans represents the shock transmission coefficient of the lower mass. TA and HA are the tibial and head accelerations. Stance is the stance time (ms).