INTRODUCTION
Experiments utilizing cadaveric models can be designed to investigate a variety of issues related to functional anatomy and orthopedic pathology. Examining isolated components of the foot or ankle is useful but may not provide meaningful insight into the structure’s overall in vivo function during activities of daily living. Recently cadaver models, consisting of the entire foot and ankle complex, have been developed in order to gain a better understanding of lower extremity biomechanics in health and disease.

This presentation focuses on describing advances incorporated into the latest iteration of a device first described by Sharkey and Hamel in 1998, the Dynamic Gait Simulator (DGS). This unique research tool was integral in completing several investigations into foot and ankle mechanics, including examinations of internal kinematics (Hamel et al., 2003), metatarsal and tibial strain in relation to cyclic overload and stress fractures (Donahue et al., 1999; Donahue et al., 2000) and bone maintenance (Peterman et al., 2001), plantar fascia function (Hamel et al, 2001; Erdemir et al., 2002), ankle trauma and reconstruction (Michelson et al., 2002), as well as issues related to the surgical management of diabetes (Sharkey and Hamel, 2000).

MOTIVATION
The DGS was designed to reproduce lower leg motion as well as the actions of the extrinsic muscles of the foot in simulations of the stance phase of walking. Despite the success of the DGS in providing researchers with a means of conducting invasive procedures on the human foot and ankle under physiologic loading conditions, model shortcomings were of concern and prompted complete redesign of the apparatus.

The motion of the knee joint in the original DGS was recreated with a cam profile that was machined into plates making modifications to the prescribed knee motion arduous and expensive. Furthermore, this lack of flexibility affected simulation fidelity and made it difficult to study specimens of varying dimensions.

In a living person, muscles that cross the knee impart a resistive moment to counteract the moments induced by ground contact, so that the joint does not freely rotate. Rotation of the simulated knee joint in the DGS had no such constraint, therefore reducing the overall fidelity of the model. In particular we noted that the minima in vertical ground reaction force that occurs in midstance was consistently lower than that measured in live subjects.

EXPERIMENTAL DESIGN
The motion of the proximal shank is recreated in the current version of the DGS with a system of three linear actuators as opposed to the cam-carriage mechanism used previously. The three actuators are programmed to recreate the 2-dimensional sagittal kinematics of the proximal fibular head and lateral maleolus during any activity that can be captured using standard gait analysis hardware and techniques. Two of
these actuators are used to recreate vertical movement of the proximal shank and constrain its angle as prescribed by data gathered in the gait lab. The third actuator reproduces the motion of the knee in the direction of travel (anterior progression in the sagittal plane).

This architecture offers the distinct advantage over the previous version by allowing researchers to investigate the effects of variations in lower leg kinematics quickly and easily. Kinematic profiles from different sized individuals under different conditions can be quickly loaded into the control algorithm to better match kinematics to specimen dimensions. This is an important improvement because of the sensitivity of internal loading to mismatches between kinematics and specimen dimensions. In addition, this feature increases the versatility of the machine as it is no longer constrained to simulate just walking. Future research is likely to investigate internal biomechanics as a function of running and jumping.

As in the original DGS, the latest version incorporates the actions of six different extrinsic muscles or muscle groups. The muscles have been grouped based on EMG measures of activation throughout the stance phase of walking. The protocol used to recreate the muscle forces incorporates independent force feedback control of the six muscle groups, as described in detail by Sharkey and Hamel in 1998. Our new iteration uses the same control strategy but the actuators used to simulate muscle contraction have been removed from the carriage assembly to reduce inertial effects and are now linked to the extrinsic tendons through cables. Control hardware has also been upgraded and improved.

RESULTS
Figure 1 compares GRFs produced by the new simulator with those generated in vivo by a subject whose lower leg kinematics were used to control the simulation.

<table>
<thead>
<tr>
<th>Force (% Body Weight)</th>
<th>% Stance Phase</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0.1</td>
</tr>
<tr>
<td>0.2</td>
<td>0.3</td>
</tr>
<tr>
<td>0.4</td>
<td>0.5</td>
</tr>
<tr>
<td>0.6</td>
<td>0.7</td>
</tr>
<tr>
<td>0.8</td>
<td>0.9</td>
</tr>
<tr>
<td>1.0</td>
<td>1.0</td>
</tr>
</tbody>
</table>

Figure 1: GRFs from a walking simulation (solid) and an in vivo walking trial (dashed).

SUMMARY
Computer controlled actuators incorporated into the latest version of the DGS have increased the versatility and fidelity of dynamic in vitro simulations of the human lower extremity.

REFERENCES

ACKNOWLEDGEMENTS
This work was partially supported by grants from NIH (R01HD3744302) and NASA (NAG9-1264)