THE ROBOTIC GAIT SIMULATOR: THE EFFECT OF EMG TO FORCE ESTIMATION

[1,2]Patrick M. Aubin and [1,3,4]William R. Ledoux
[1]VA RR&D Center of Excellence for Limb Loss Prevention and Prosthetic Engineering, Seattle, WA 98108, and Departments of [2]Electrical and [3]Mechanical Engineering, and [4]Orthopaedics & Sports Medicine, University of Washington, Seattle, WA 98195
email: wrledoux@u.washington.edu, web: http://www.amputation.research.va.gov/

INTRODUCTION

Cadaveric models are useful to further our understanding of foot and ankle function, biomechanics, and pathology. There are significant challenges however in accurately recreating *in vitro* the complex foot and ankle kinematics, musculotendinous forces, and ground reaction forces that occur during gait. To address these challenges we have developed the robotic gait simulator (RGS) by employing a 6-DOF parallel robot and nine force control tendon actuators. The inputs into the RGS are the stance phase tibial kinematics, electromyography (EMG) of nine extrinsic lower limb muscles, and the vertical ground reaction force (vGRF). We created a set of heuristics to manually adjust the vGRF until it matches normative *in vivo* gait data. Results from three cadaveric specimens each performing three gait trials are presented. Comparing the temporal characteristics of the *in vitro* and *in vivo* Achilles tendon force and vGRF reveals the limitations of a common EMG to force estimation method.

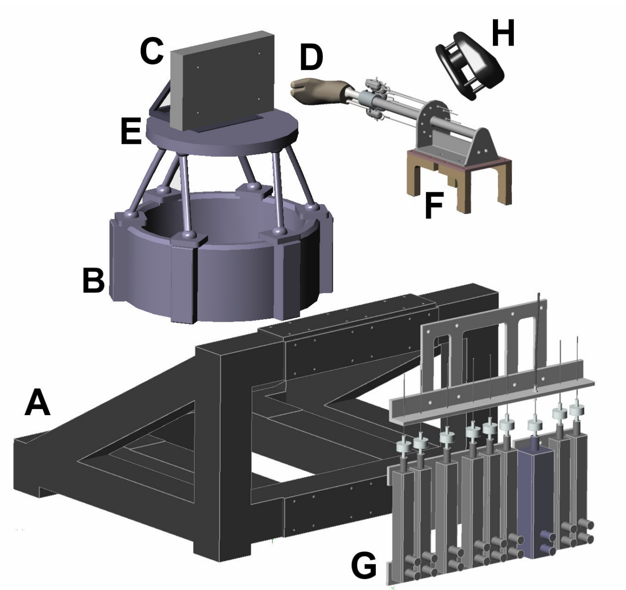

Figure 1: An exploded view of the Robotic Gait Simulator (RGS) with surrounding frame (A), the R2000 (B), mobile force plate (C), cadaveric foot (D), mobile platform (E), tibia mounting frame (F), tendon actuation system (G) and motion analysis system (H).

METHODS

Ten healthy subjects were recruited and asked to perform five gait trails each after providing informed consent for this Institutional Review Board approved study. A 12-camera Vicon motion analysis system sampled kinematic data at 250Hz while a Bertec force plate sampled kinetic data at 1500Hz.

The RGS consists of an R2000 parallel robot with a mobile Kistler force plate, nine force control tendon actuators and load cells in series with each tendon, a tibia mounting frame, and a six-camera Vicon motion analysis system (Figure 1). To simulate gait the tibia is fixed while the R2000 moves the force plate, i.e. the "ground," to recreate the relative tibia to ground motion.

The force requirements for the nine extrinsic tendons were estimated from each muscle's physiological cross sectional area (*PCSA* cm^2), the maximum specific isometric tension (*MST* N/cm^2), the EMG activity during gait (EMG % of maximum voluntary contraction) [1] and a gain (*G*) (eq. 1).

$$F_T = G \cdot PCSA \cdot MST \cdot EMG \qquad \text{eq. 1}$$

Three fresh frozen cadaveric right feet each performed three trials. The simulation from heel strike to toe off was scaled to $1/15^{th}$ of physiological velocity. The vGRF and non-Achilles tendon forces were scaled to one half of the *in vivo* forces to minimize the chances of failing frail cadaveric specimens. During the simulation the vGRF and tendon forces were recorded at 1000Hz.

In order to achieve the desired vGRF two parameters were manually adjusted iteratively between gait simulations, namely, the trajectory of the mobile force plate along the superior-inferior tibial axis and the Achilles tendon gain, *G* (eq. 1). The superior-inferior offset was used to affect the first peak of the vGRF while adjustments to the Achilles tendon gain G were used to affect the second peak of the vGRF. Gait simulations were repeated iteratively while manually tuning these two parameters until the magnitude of the first and second peak of the simulated vGRF approximated the *in vivo* vGRF peaks. No attempt was made to control the temporal characteristics.

RESULTS AND DISCUSSION

The vGRF from the nine *in vitro* gait simulations reached peak magnitudes similar to the *in vivo* vGRF scaled to one half body weight (0.5BW) (Figure 2). The average % error across all trials between the *in vivo* and *in vitro* vGRF was 9.4% for the first peak and 3.1% for the second peak. The average RMS errors across all trials between the target and actual *in vitro* tendon force for the eight non-Achilles extrinsic tendons were between 2.8N and 8.5N RMS (Table 1). The Achilles tendon had the highest average RMS error at 20.3N, but its RMS error as a percent of peak force was the smallest at 2.3% (Table 1).

Inspection of the temporal characteristics of the *in vitro* vGRF and Achilles tendon force during late stance demonstrated some of the limitations of our *in vivo* muscle force estimate based on EMG. After heel rise the Achilles tendon force strongly dictates the vGRF. The early decrease in the *in vitro* vGRF indicates an underestimation of Achilles tendon force between 75% and 100% of the stance phase. This is confirmed by comparing our estimated Achilles tendon force derived from EMG measurements to those obtained *in vivo* using a fiber optic measurement technique [2] (Figure 2). Ishikawa *et al.*'s *in vivo* measurement of Achilles tendon force has a peak occurring at approximately 75% of stance phase compared to a peak at approximately 65% for the Achilles tendon force estimate derived from EMG (Figure 2). Our EMG-scaled model (eq. 1) used to estimate muscle force does not include muscle activation or deactivation first order dynamics, nor does it account for the parallel passive element present in a Hill-based muscle model. Including the electromechanical delay (EMD) into our model would delay the time at which peak Achilles tendon force occurs and the rate at which it decreases in late stance. Similarly, passive force developed in response to a lengthened muscle-tendon unit exists late in the stance phase [2]. Bogey *et al.*'s EMG to force estimation method which models the passive force production and EMD of soleus and gastrocnemius has temporal characteristics similar to the *in vivo* force data [3]. These findings have motivated us to use *in vivo* measurements of Achilles tendon force or more sophisticated EMG to force models to estimate the target tendon forces for future RGS simulations.

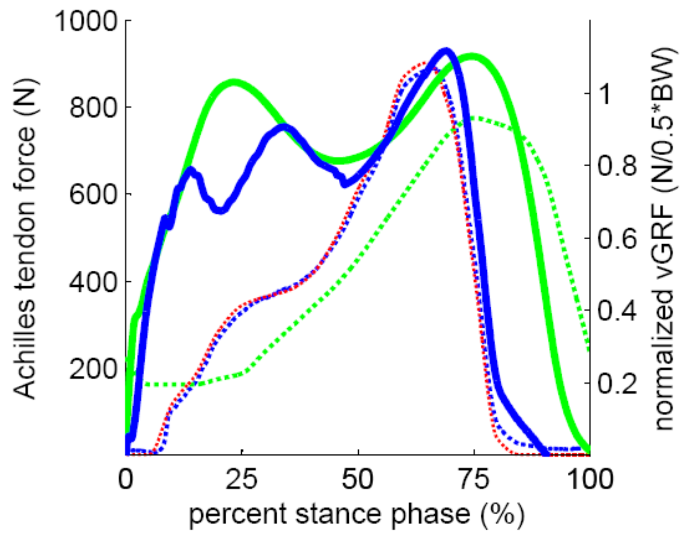

Figure 2: *In vitro* simulation results compared to *in vivo* gait data. The thick solid lines are mean *in vivo* (green) and *in vitro* (blue) vGRF (N/0.5BW). Mean Achilles tendon forces (N) are shown as the thin dotted lines with the EMG based estimate in red, *in vitro* simulation results in blue and Ishikawa *et al.*'s *in vivo* fiber optic measurement in green.

REFERENCES

1. Blackman, A.J. *et al.*, JOR, in revision.
2. Ishikawa, M. *et al.*, J Appl Physiol 99, 603-8 (2005).
3.Bogey, R.A *et al.*, IEEE Trans Neural Syst Rehabil Eng 13, 302-10 (2005).

ACKNOWLEDGEMENTS

This research was supported in part by the VA RR&D, grant A3923.

Table 1: Tendon force tracking RMS error expresses as newtons (N) and percent of peak force (%).

	Ach	EDL	EHL	PL	PB	FHL	FDL	TP	TA
N	20.3	4.8	4.7	5.6	5.0	7.8	2.8	8.5	6.8
%	2.3	16.4	35.7	11.7	27.9	15.1	13.4	11.4	10.9