Development of Muscle Actuated Robotic System for the Investigation of Lower Extremity Function

W. A. Spivey[1], C. M. O'Cain[1], B. Gepner[1], E. Spratley[1], and J. Kerrigan[1]
[1]Center for Applied Biomechanics, University of Virginia;

ABSTRACT

Prior to automotive impact, occupants often engage their muscles to brace themselves for collision, and despite this fact, few injury prediction tools take into account the effect of active musculature during injury. This study details the development and initial testing of a Muscle Actuated Robotic System (MARS), which could eventually be used to examine the effect of musculature on injury. The proof of concept study for MARS is to investigate the effect of body weight on bony kinematics in the foot during walking. Development of the MARS included coupling both the hardware and control systems of a 6-degree of freedom position and force/torque controlled serial robot with a set of nine linear actuators. Results show the MARS is capable of replicating human-like gait, reproducing the target tibia kinematics within 1 mm and 0.1 degree respectively, and that both, the calcaneus rolls, and the midfoot arch drop increased with the increase of input body weight. Although the current study evaluates gait, the robotic system is capable of applying muscle forces throughout various joints in the body.

INTRODUCTION

Prior to automotive impact, occupants often engage their muscles to brace themselves for collision (Hault-Dubrelle, 2011). This is common in frontal crashes, where lower extremity injuries remain the most frequently injured body region (Ye, 2015). Despite this typical occupant behavior, few injury prediction tools take into account the effect of active musculature during injury. Computational human body models (i.e. Active THUMS and ActiveHuman) are beginning to include active musculature; however, validation data is currently limited to volunteer data. Biomechanics research on living subjects is limited to what can be measured non-invasively; consequently, injury cannot be systematically evaluated in volunteers. Post-mortem human surrogates (PMHS) remain the gold standard for human injury measures. However, PMHS lack muscle activation, which has been shown to effect injury tolerance (Funk, 2002). Simulating the effects of active musculature in the lower extremity would allow for examining the bony dynamics that occur during gait or other natural loading scenarios. This study details the development and initial testing of a Dynamic Muscle Activation System (MARS), which could eventually be used to examine the effect of musculature on injury. This study aims to develop a method to apply active musculature to PMHS leg/ankle/foot specimens that includes parallel controllers to adjust muscle forces and leg kinematics to produce biofidelic loading scenarios. This methodology will be essential in examining how muscle loads affect injury risk and bony motion in the foot and ankle. In this study, MARS was applied to tune a PMHS specimen to simulate quarter to full body-weight gait, and to compare the test specimen responses to gait dynamics data captured in volunteer testing.

METHODS

Part I – MARS Design

Development of the MARS included coupling both the hardware and control systems of a 6-degree of freedom (6-DoF) position and force/torque controlled serial robot (Kuka KR300 R2500 Ultra, Kuka Robotics Corporation, Augsburg, Germany) with a set of nine linear actuators (Kollmorgen Electric Cylinder (EC) 3 and 4 Series, Kollmorgen Corporation, Radford, VA). The Kuka robot is capable of carrying a 300 kg payload 2496 mm away from the base of the robot with an accuracy of 0.06 mm. The smaller actuators (EC3) are capable of generating 1500 N at a maximum speed of 533.4 mm/s, while the larger actuator (EC4) is capable of 4000 N at a maximum speed of 533.4 mm/s. Once coupled, the system has the power, stiffness, and operating rate necessary to simulate human gait under realistic loading scenarios.

In order remove the complication of the actuators, being mounted stationary while the robot is moving, the actuators were mounted to the robot (Figure 1). Each actuator was connected to a tendon via a steel cabling system. This system was comprised of the steel cable surrounded in a low friction cable housing connected to a load cell (Honeywell Model 31, Honeywell, Charlotte, NC) and then the tendon. The nine actuators were connected to the following tendons: Achilles, tibialis anterior, tibialis posterior, flexor hallucis longus, flexor digitorum longus, extensor hallucis longus, extensor digitorum longus, peroneus longus, and peroneus brevis. These tendons represent the largest contributors to foot and ankle motion (Perry, 1992). The cable housing allowed for the routing of the steel cable while the robot is moving without the need for a pulley system. By coupling these systems, the robot was able to control tibia kinematics, while the linear actuators can control individual foot muscles.

Both the robot and actuators are controlled using a software package (simVitro, Cleveland Clinic BioRobotics, Cleveland, OH) capable hardware integration and communication, rapid data collection and processing. The software package has a library of transformations based major joint areas (ex. spine, knee, foot) for replicating biofidelic motions. All of the MARS hardware components include: the robot's actuators and load cell, tendon linear actuators and load cells, and a coordinate digitization arm are interconnected and can actively communicate through simVitro. The software allows for the simultaneous control of tibia kinematics, ground reaction forces, and muscle forces. Each simVitro module is capable of handling a different joint in the body, defined by coordinates that are digitized, for instance the foot and ankle module was used with the MARS, requiring digitization of the International Society of Biomechanics (ISB) definition of the foot coordinate system (Wu, 2002). The robot is capable of both position and force control, from encoders on the robot's motors, and a control load cell mounted at either the end effector of the robot or in a platform with dynamic gravity transformation included when necessary. The actuators are only capable of force control, with the control load cells mounted in between the actuator and tendon along a guide ring to keep the load cell from swinging during gait.

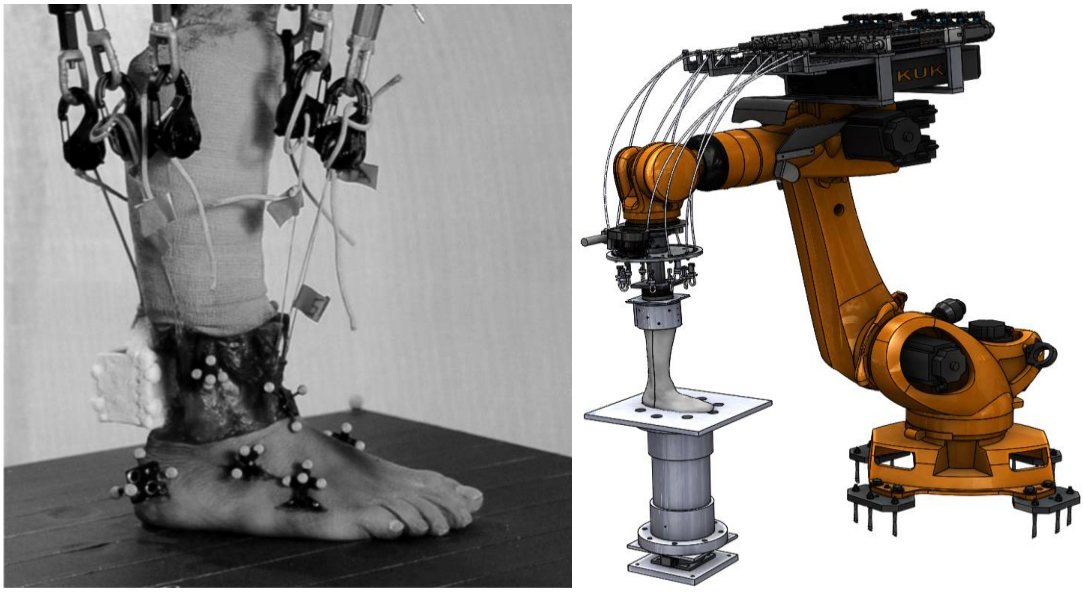

Figure 1: Left –Photo of instrumented PMHS on the MARS. Right – Rendering of MARS.

Part II – Data Collection and Processing

Data for the tests are collected in two ways, kinematics and kinetics of the robot, as well as bony kinematics from motion tracking arrays. The simVitro software records kinematics and kinetics of the robot, transformed to a specified coordinate frame, in this case the ISB foot definition. The robot kinematics and kinetics were processed by using a low pass 1 Hz Butterworth phaseless filter.

In order to capture kinematics in the hind- and mid-foot, 3D motion tracking arrays were rigidly affixed to the following eight bones: tibia, fibula, calcaneus, talus, navicular, cuboid, first metatarsal, and fifth metatarsal. Marker arrays were digitized from a pretest CT scan, along with their position relative to the foot bones in order to establish a rigid body transformation between marker arrays and their respective bones. To allow for consistent starting coordinate systems, an anatomic coordinate frame was created from landmarks on the tibia, medial and lateral tibial plafond, and medial and lateral malleolus, and then transformed to the centroid of each of the other bones. During testing, a laboratory reference frame was established using motion-tracking targets attached to the plane of the walking platform. All of the motion capture data is reported in this lab reference frame, with the anterior direction being +X, medial direction being +Y, and superior direction being +Z. Using the established transformations, motion-tracking trajectories are applied to all of the marker arrays and their bones, respectively.

Part III – Gait Modeling using MARS

A single male PMHS (46 years, 99.3 kg, 175.3 cm) lower extremity was tested to evaluate the accuracy of the system. The donations were obtained and treated in accordance with the ethical guidelines established by the United States National Highway Traffic Safety Administration (NHTSA), and all testing and handling procedures were reviewed and approved by an institutional review board for human surrogate use at the University of Virginia. Approximately 100 mm of tissue was removed circumferentially around the ankle to expose the tendons of the muscles of the leg. Eight of the nine actuators were then affixed to tendons using polyester surgical thread with a Krakow stitch, used typical for tendon reconstruction surgeries. Due to the greater magnitude of force required, the Achilles was clamped using a custom cryoclamp, and was frozen. During testing, tendon force was generated by displacement in the linear actuators using force control feedback. Force time histories for all nine tendons were replicated from literature. Briefly, those studies used dynamometers and electromyography to ascertain maximum voluntary contraction (Perry, 1992) coupled with physiologic cross sectional areas of each tendon (Wickiewicz, 1983) to estimate the level of muscle activation force. Tibial kinematics were imposed by the robot through displacement control. The tibial kinematics were recorded via motion tracking markers on a volunteer in a gait lab, and simultaneously the ground reaction forces were recorded during the gait trials.

Four body weight conditions were investigated: 25%, 50%, 75%, and 100% body weight (BW). For each BW, a target ground reaction force (GRF) time-history was used for optimize the tibialis anterior force during heel strike, z-position of the tibia during midstance, and Achilles force during toe-off. Both GRF and muscle time histories were scaled linearly for the BW conditions. Each BW condition had 8-12 optimization runs, with three data collection runs each. The optimization routine consists of a linear reduction of error, using the calculated error between the recorded signal and desired signal and weighing factor to scale a new input response.

RESULTS

Four BW conditions were investigated to determine the effects of different loading conditions on foot bone response. Prior literature has shown that the vertical GRF curves of human gait are characteristically bimodal, with the first maxima occurring due to the heel striking the ground and the second peak when the plantar flexing foot causes the toes to push off (Umberger, 2007). Through optimization during the one BW condition, the system was able to reliably reproduce the target input, within 5% error, and nominally bring heel strike close to the target, within 15% (Figure 2). Other BW conditions performed similarly, with the greatest amount of error occurring during heel strike, and the least amount of error during toe off. Tibia kinematics were likewise very repeatable with position and angles within 1 mm and 0.1 degree respectively (Figure 3).

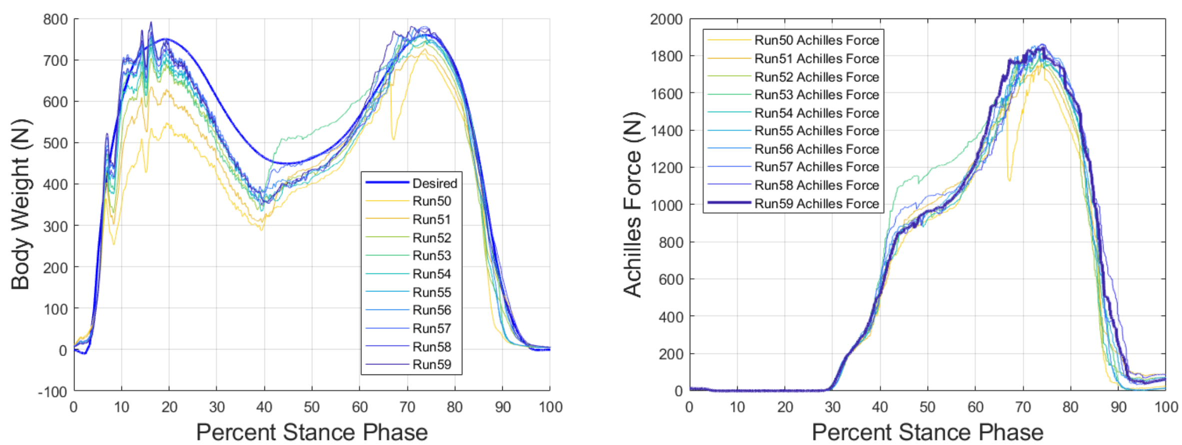

Figure 2: Left- Target GRF vs Measured GRF for multiple optimization runs vs time in 100% BW condition. Right - Demonstration of optimization routine showing Achilles force vs time.

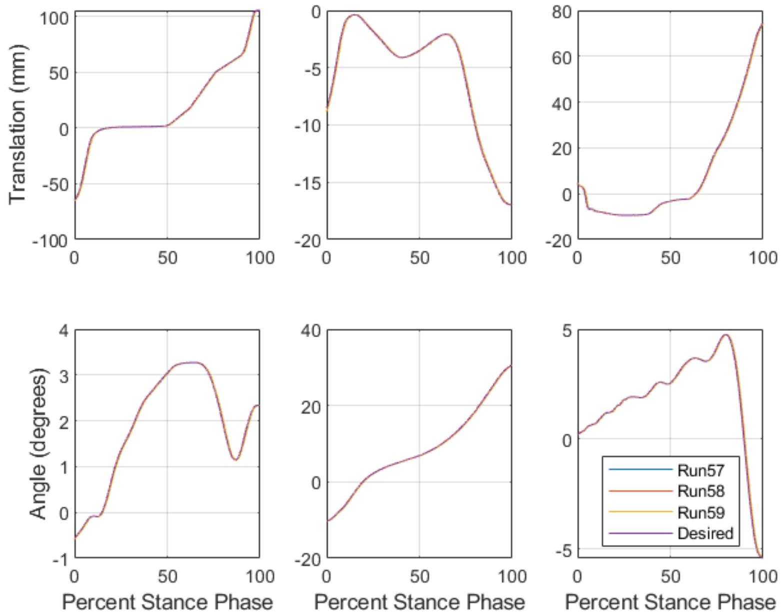

Figure 3: Target Kinematics vs Measured Kinematics in 100% BW condition; Scale is too large to see deviation in the curves; Top row: X-translation, Y-translation, Z-translation, Bottom row: X-rotation, Y-rotation, Z-translation

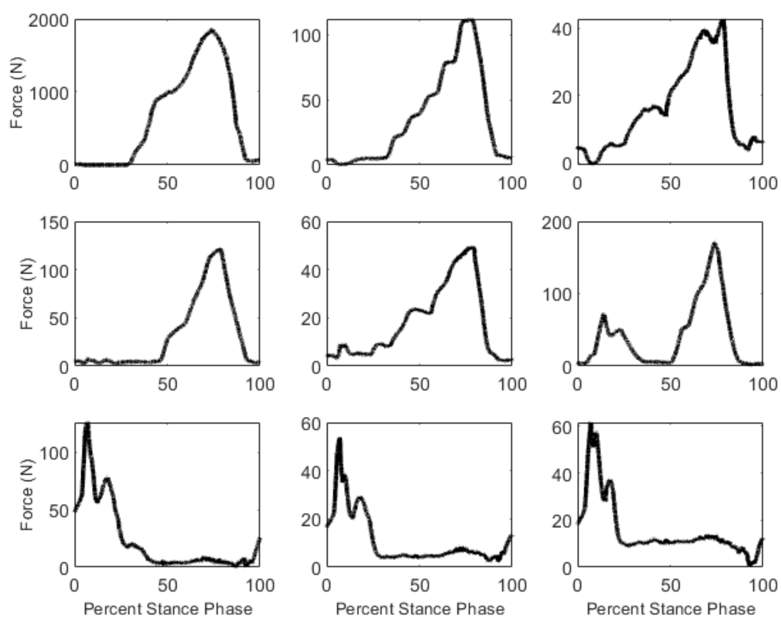

Figure 4: Muscle Force Time Histories for 100% BW condition; Top row: Achilles, peroneus longus, peroneus brevis, Middle row: flexor hallucis longus, flexor digitorum longus, tibialis posterior, Bottom row: tibialis anterior, extensor digitorum longus, extensor hallucis longus

Increasing BW led to changes in global foot kinematics. The calcaneus rotated 6 degrees about the x-axis, known as eversion, under 100% BW versus 2 degrees under 25% BW. Additionally, the navicular was shown to translate 18 mm along the z-axis towards the ground during the 100% BW condition, while only translating 15 mm during 25% BW (Figure 5).

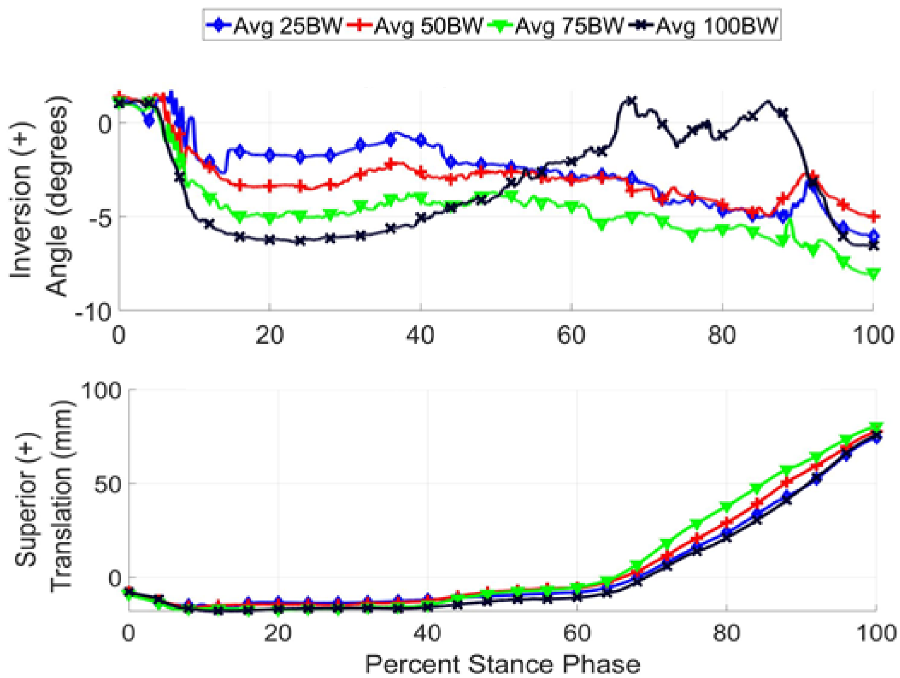

Figure 5: Change in hindfoot inversion/eversion as measured by rotation about calcaneus x-axis. Drop in longitudinal arch height as measured by Z-translation of the navicular

DISCUSSION

The system was able to reproduce gait-like motions of a PMHS lower extremity. Input tibial kinematics were matched closely and with little variance between trials. A small phase shift was observed for all trials due to the control system. Trends noticed during testing, notably navicular height are consistent with those reported for volunteer subjects and are attributable to the longitudinal arch spreading and compressing with larger amounts of axial load (Lundgren, 2008). However, more specimens are necessary to determine if the reported bony kinematics are trends as the conducted experiment only have a sample size of one. Even with this limitation, the study demonstrates a proof of concept for the testing methods and design of the system itself

The 100% BW trial deviated during toe off from the expected trend; this is primarily due to the foot sliding posteriorly, suggesting that the imposed forces in the other directions is not sufficient. Further improvements can be made to the optimization routine to include these other directions. Additionally, other muscle forces could be added to the optimization routine as only two of the nine muscles are being optimized. The optimization algorithm currently only acts in three major events: heel strike, midstance, and toe off, but could be expanded to allow different muscles to overlap multiple regions.

While this study evaluates gait, there is no hardware limitation on attaching the motors to tendons in a different body region. As long as the software package supports the desired joint, and analogs inputs, in this case volunteer recorded vertical GRF can be attained, the method should apply.

CONCLUSIONS

Results demonstrate the system was able to generate kinetic data (ground reaction force) and kinematic data (calcaneus and navicular motions) consistent with published volunteer response. Although the current study evaluates gait, the robotic system was developed to generically apply muscle forces throughout the body. Preliminary results of this study suggest that the MARS can generate a combination of biofidelic muscle forces, bony motion, and reaction forces in a closed loop parallel control to use cadaveric tissue to represent the loads and motions occurring in the human body.

ACKNOWLEDGEMENTS

Funding for the project was provided by the University of Virginia through the Research and Innovation Award program and the Engineering in Medicine initiative.

REFERENCES

FUNK J., CRANDALL J., TOURRET L., et al. The Axial Injury Tolerance of the Human Foot/Ankle Complex and the Effect of Achilles Tension. ASME. J Biomech Eng. 2002;124(6):750-757. doi:10.1115/1.1514675.

HAULT-DUBRULLE, A., ROBACHE, F., DRAZETIC, P., GUILLEMOT, H., MORVAN, H., Determination of pre-impact occupant postures and analysis of consequences on injury outcome—Part II: Biomechanical study, Accident Analysis & Prevention, Volume 43, Issue 1, 2011, Pages 75-81, ISSN 0001-4575.

LUNDGREN, P., NESTER, C., LIU, A., ARNDT,A., JONES, R., STACOFF, A., WOLF, P., LUNDBERG, A., Invasive in vivo measurement of rear-, mid- and forefoot motion during walking, Gait & Posture, Volume 28, Issue 1, 2008, Pages 93-100, ISSN 0966-6362.

PERRY, J., Gait analysis, Thorofare, NJ: Slack Incorporated, 1992.

UMBERGER, B., MARTIN, P. Mechanical power and efficiency of level walking with different stride rates. Journal of Experimental Biology. 2007 210: 3255-3265; doi: 10.1242/jeb.000950.

WICKIEWICZ T., ROY R., POWELL P., EDGERTON V. Muscle architecture of the human lower limb. Clin Orthop Relat Res. 1983 Oct 275-283. PMID: 6617027.

WU, G., SIEGLER, S., ALLARD, P., KIRTLEY, C., LEARDINI, A., ROSENBAUM, D., WHITTLE, M., D'LIMA, D., CRISTOFOLINI, L., WITTE, H., SCHMID, O., STOKES, I., ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion—part I: ankle, hip, and spine, Journal of Biomechanics, Volume 35, Issue 4, 2002, Pages 543-548, ISSN 0021-9290.

YE, X., POPLIN, G., BOSE, D., FORBES, A., HURWITZ, S., SHAW, G., CRANDALL, J., Analysis of crash parameters and driver characteristics associated with lower limb injury, Accident Analysis & Prevention, Volume 83, 2015, Pages 37-46, ISSN 0001-4575.