Development of a Muscle Actuated Robotic System (MARS) for the Investigation of Lower Extremity Bony Kinematics

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I. INTRODUCTION

Prior to automotive impact, occupants often engage their muscles to brace themselves for collision [1]. This is common in frontal crashes, where lower extremity injuries remain the most frequently injured body region [2]. Despite this typical occupant behaviour, few injury prediction tools take into account the effect of active musculature during injury. Computational human body models [3], i.e. Active THUMS and ActiveHuman, are beginning to include active musculature; however, validation data is currently limited to volunteer data. While volunteer data is useful, biomechanics research on living subjects is limited to what can be measured non-invasively and non-injuriously. Consequently, injurious loading rates cannot be systematically evaluated in volunteers. Post-mortem human surrogates (PMHSs) remain the gold standard model for human injury prediction. However, PMHSs lack muscle activation, and it has been shown that applying muscle tension in PMHS models can affect injury tolerance [4]. Simulating the effects of active musculature in the lower extremity would allow for examining the bony dynamics that occur during body loading scenarios. This study details the development and initial testing of a Muscle Actuated Robotic System (MARS), and aims to develop a methodology to apply active musculature to PMHS leg/ankle/foot specimens to address how that musculature effects injury tolerance and bony kinematics.

II. METHODS

The major components of the MARS are a serial robot, tendon actuators, coordinate digitising arm, motion analysis system, and a six degree-of-freedom load cell (6-DoF) (Fig 1A).

The 6-DoF position and force/torque controlled serial robotic test system was used to drive previously collected *in vivo* tibia kinematics. Gait was used as the test data set due to the wealth of publicly available volunteer data [5-6]. The PMHS specimen was affixed to the end-effector of the robot with a rigid urethane potting of the proximal tibia. Prior to testing, multiple points on the PMHS limb were digitised and used to generate a specimen-specific ankle coordinate system with respect to the robot's coordinate system. Nine linear actuators were used to pull tension in each of the following nine tendons: achilles, tibialis anterior, tibialis posterior, flexor hallucis longus, flexor digitorum longus, extensor hallucis longus, extensor digitorum longus, peroneus longus, and peroneus brevis. Muscle force time-histories were generated using percent activation and muscle physiological cross sectional area volunteer data [7-8].

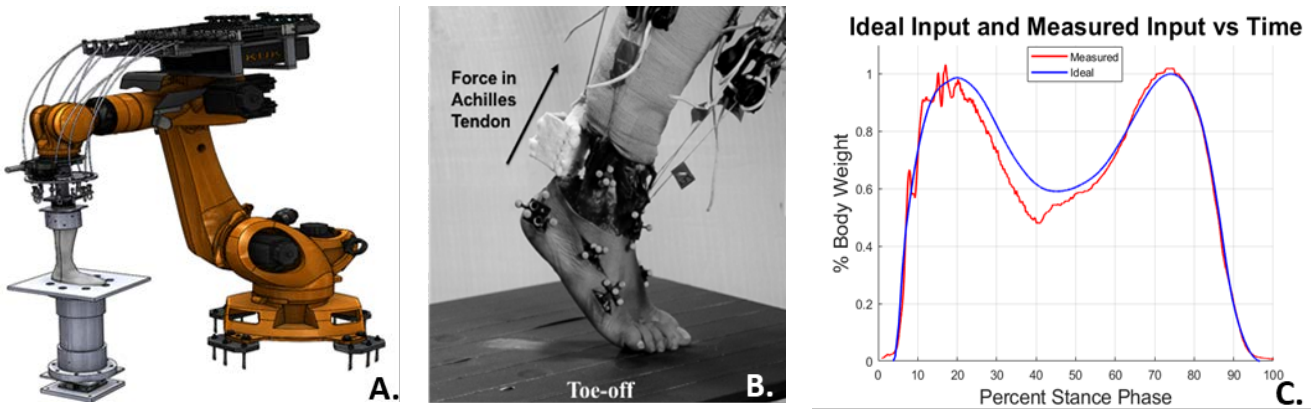

Fig 1. (A) Rendering of test setup (B) photo during testing (C) sample input curve.

The single PMHS left lower extremity used for this study was disarticulated at the knee (male, 46 years, 99.3 kg, 175.3 cm). Approximately 150 mm of the proximal tibia was denuded to allow for connection to the end of the robot. Additionally, a 50 mm window of skin was removed from around the circumference of the ankle just

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above the medial malleolus in order to expose the tendons of each of the extrinsic muscles inserting in foot. Each actuator was then affixed to its respective tendon using polyester surgical thread by means of a Krakow stitch. Due to the greater tension magnitude required, the Achilles tendon was attached to its actuator using a custom aluminum cryoclamp. Retroreflective 6-DoF motion tracking arrays were rigidly screwed into to the following bones: tibia, fibula, calcaneus, talus, navicular, cuboid, first metatarsal, and fifth metatarsal.

Four rates of gait were evaluated: one gait cycle over 20 seconds, 10 seconds, 6 seconds, and 4 seconds. A cycle is defined here as the time from heel strike to toe off. All trials were conducted at one-quarter body weight. For each cycle, a target ground reaction force (GRF) time-history was used to optimise the tibialis anterior force during heel strike, superior position of the tibia during midstance, and Achilles force during toe-off. Each condition had 8-12 optimisation runs, with three bony kinematic data collection runs each.

III. INITIAL FINDINGS

Time between heel strike and toe off, known as stance phase, was varied to determine the effects of different speeds on foot bone response. The system was able to trace the GRF approximately within 15% of the input target force (Fig 1C). Two initial measures have been observed, calcaneus eversion and navicular Z-translation. No major differences were observed in calcaneus eversion, with the 20-second trial everting 3 degrees, and the 4-second trial everting 2 degrees. Navicular Z-translation was different between trials. The minimum negative navicular translation occurred in the 20-second trial (20 mm), and the maximum occurred 4-second trial (28 mm).

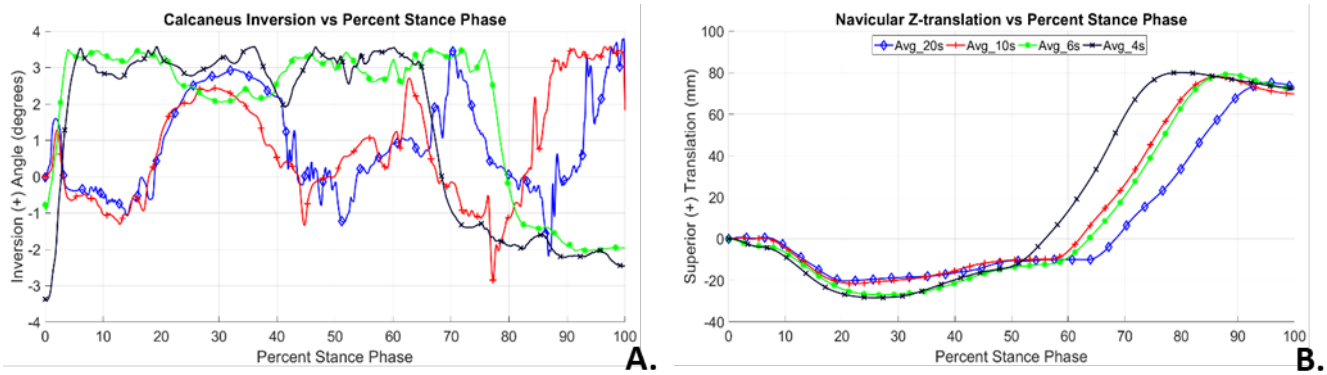

Fig 2. (A) Hindfoot eversion as measured by rotation about calcaneus x-axis. (B) Z-translation of the navicular.

IV. DISCUSSION

The ability of MARS to match volunteer GRF and muscle force time history demonstrates the feasibility of adding biofidelic musculature effects into PMHS testing. Although the current study evaluates gait, the robotic system was developed to generically apply muscle forces throughout the body. Therefore, MARS lends itself to the evaluation of muscle activation effects with PMHSs in multiple body regions.

Specific bony kinematic results of this study show as rate of applied load increased, the magnitude of navicular downward motion also increased. These results suggest that the longitudinal arch of the foot flattens to a higher degree as the rate of applied load increases. This is not surprising though due to the naturally viscoelastic behaviour of biological tissue. The change in foot arch demonstrates the bony kinematics and resulting load distribution can be affected by rate, even when active musculature is applied. However, calcaneus rotation did not change based on the rate, suggesting that calcaneus motion has a higher dependence on magnitude of loading as compared to rate of loading in a gait scenario. It should be noted that results of this study are limited by the rates used in testing. Future work is required to evaluate trends in bony kinematics at higher rates at which injuries typically occur. While injury was not generated in the current study, preliminary results suggest that MARS can repeatedly generate a combination of biofidelic muscle forces, bony motion, and reaction forces that are all integral components to understanding how muscle activation could affect injury.

V. REFERENCES

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