EVIDENCE OF ISOMETRIC FUNCTION OF THE FLEXOR HALLUCIS LONGUS MUSCLE IN NORMAL GAIT

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INTRODUCTION
Biarticular and multiarticular muscles are known to possess unique biomechanical and functional characteristics. The activation patterns of multiarticular muscles in multijoint movements have been studied in the past. These muscles are thought to function as ‘nearly isometric structures’ transferring mechanical energy from one joint to another rather than doing substantial work themselves during certain movements. (Bobbert and van Ingen Schenau 1988). However, the theory of ‘isometric function of multiarticular muscles’ is based on anatomical and geometric considerations, indirect force estimates and electro-myographic (EMG) data. Direct experimental data supporting isometric function in humans are scarce.

The Flexor Hallucis Longus (FHL), a multiarticular muscle, is the major great toe flexor. Originating from the distal third of the fibula and traveling across the entire length of the sole, FHL inserts on the base of the distal phalanx of the great toe. The FHL tendon crosses multiple joints along its path, notably the ankle and the 1st metatarso-phalangeal (MTP) joint. Neural drive to FHL is maximal during the push-off phase of the gait cycle. The dynamic coupling of the 1st MTP and ankle joints becomes apparent during the early push-off phase when the former is dorsiflexing while the latter is plantarflexing.

We hypothesized that the FHL muscle operates isometrically during the gait cycle, and that, the constant length that is maintained is specific to each individual foot.

METHODS
Using a custom-made robotic device, called the robotic dynamic activity simulator (RDAS), we created dynamic simulations of the stance phase of walking gait in non-embalmed human cadaver feet. The capacity of RDAS to simulate walking has been previously validated (Sharkey and Hamel, 1998; Hoskins, 2004). Kinematic data of the proximal shank as measured in live subjects during the stance phase of normal, unshod walking, and EMG data (Perry et al., 1992), scaled according to relative muscle cross-sectional area, were used as input for the simulations. Lower extremity specimens with intact tendons were mounted in the RDAS and were loaded under near physiological conditions, i.e. peak vertical ground reaction forces (GRF) of 500 N. A set of three linear actuators was used to recreate the pre-recorded kinematics of the proximal shank. Another set of actuators, linked to the musculo-tendinous junctions via cables and cryogenic clamps, simultaneously prescribed the muscle activity (i.e. ‘target’ muscle force profiles derived from the scaled EMG data). The three components of the GRF, as well as displacements and forces of the major extrinsic tendons of the foot were dynamically recorded over each simulation.

During initial trials, predetermined force profiles for individual tendon units were achieved using force feedback (FFB) control while proximal tendon displacements were recorded. In subsequent trials, FFB control was used for all tendon units except the FHL, which was controlled using displacement feedback (DFB) with the
forces generated in the tendon being monitored. Constant FHL lengths at different positions were prescribed as input. Successive trials were aimed at finding the specific (i.e. ‘neutral’) position at which the recorded FHL force profile matched its ‘target’ force profile as estimated from EMG and anthropometric measures. Subsequently, constant FHL positions 2, 4, 6 and 8 mm proximal to the ‘neutral’ position were prescribed in successive trials to monitor the corresponding changes in the forces generated in the FHL tendon (fig.1).

RESULTS AND DISCUSSION
The total excursion of the FHL tendon during foot movements alone, without any toe movement, is reported to be 27 mm (Hintermann, 1994); however, we found it to be 6-8 mm on an average, during walking simulations under FFB control. The recorded FHL force profiles under FFB control were identical to the target profiles. Likewise, at the ‘neutral position’ under DFB control, the FHL force profiles were comparable to the target force profiles. The neutral position was found to be within the middle of the FHL excursion range and was specific for each specimen. Furthermore, setting the FHL at constant positions located 2, 4, 6 and 8 mm proximal to the defined neutral position produced dramatic and proportional increments in the peak tendon forces. These findings have potentially interesting implications for the control of locomotion. The central neural drive to FHL possibly seeks to maintain a specific length over the gait cycle and thus contributes to the coupled dynamics between the ankle and 1st MTP joints. While central pattern generators (Yamaguchi, 2004) seem to regulate the reciprocal activation of the proximal limb muscles during locomotion, some distal muscles could likely be controlled via length-servo mechanisms. The FHL appears to be one such muscle. The negative feedback provided by the stretch reflex circuit (Matthews, 2004) seems ideally suited for such control; and has previously been postulated to be involved in the control of limb impedance (Granata, 2004). We suggest that locomotion control might employ two mechanisms: reciprocal muscle activations via central pattern generators, and impedance control via length-servo control. This idea has potential implications for the conditions of impaired stretch reflex function, e.g. peripheral neuropathies.

CONCLUSION
FHL muscle can produce physiological tendon forces during ambulation while operating isometrically.

REFERENCES
Hintermann et al. (1994) Foot & Ankle International Jul;15(7):386-95