DESIGN OF TWO LIGHTWEIGHT HIGH BANDWIDTH, TORQUE-CONTROLLED ANKLE EXOSKELETONS

Kirby Ann Witte and Steven H. Collins

Carnegie Mellon University, Pittsburgh, PA, USA
Email: kwitte@andrew.cmu.edu, web: http://biomechatronics.cit.cmu.edu

INTRODUCTION

Lower-limb exoskeletons have the potential to enhance rehabilitation [1], assist walking for people with gait impairments, reduce the metabolic cost of normal and load-bearing walking [2], improve stability and probe interesting questions about human locomotion. Designing effective lower-limb exoskeletons may be simplified by focusing on a single joint. During walking, the ankle produces larger peak torques and performs more positive work than either the hip or the knee [3], thereby making it an effective location for assistance.

The ankle joint experiences a wide range of velocities during walking, with plantarflexion occurring rapidly. It is therefore desirable to have high bandwidth and high torque capabilities to accurately produce a wide variety of torque profiles.

Many exoskeletons have been developed employing different approaches to design, actuation and control, but the most effective mechanical methods of assisting the ankle remain unclear. The process of developing and testing our devices has produced several guiding principles of exoskeleton design that may influence future exoskeleton development.

METHODS

We designed, built and tested two ankle exoskeletons, here called Alpha and Beta, for use with a tethered emulator system (Fig 1 A) described in detail in [5]. Both exoskeletons feature strong lightweight frames, comfortable three-point contact with the leg, series elastic elements for improved torque control [4], modularity, structural compliance in selected directions, and sensors for measuring torque, joint angle, and phase of gait. Torque is controlled using a combination of proportional feedback, damping injection and iterative learning during walking tests. Series elasticity is provided by leaf springs in the Alpha design and a coil spring in the Beta design. Spring type strongly affected the overall exoskeleton envelopes.

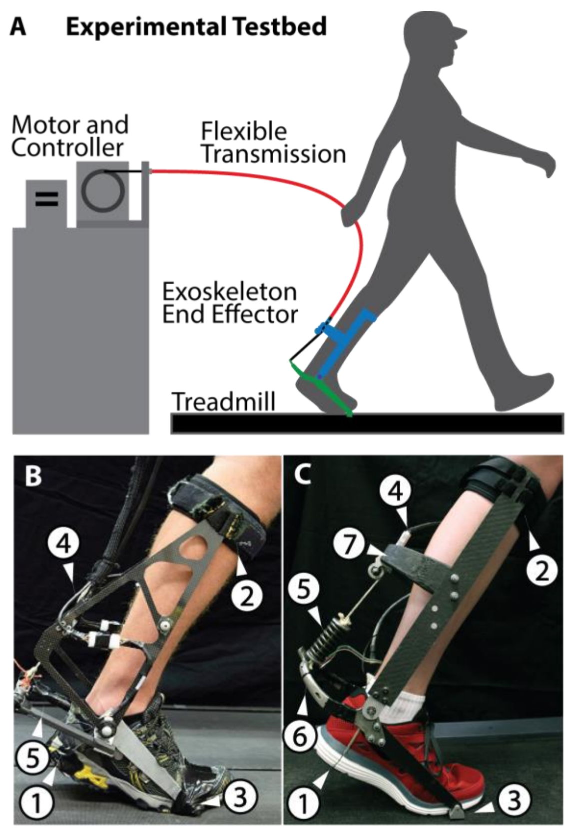

Figure 1: Emulator system and exoskeletons. **A** The testbed comprised a powerful off-board motor and controller, a flexible transmission, and an ankle exoskeleton end-effector. **B** The Alpha design. Each exoskeleton contacted (1) the heel with a string, (2) the shin using a strap, and (3) the ground with a hinged plate embedded in the shoe. The Bowden cable conduit attached to (4) the shank frame, while the Bowden cable terminated at (5) the series spring. **C** The Beta design. In addition to (1-5), this device has (6) a titanium ankle lever made using direct-metal additive manufacturing and (7) a tubular carbon fiber Bowden cable support.

The Alpha device (Fig. 1B) was designed to provide compliance in selected directions. The Beta exoskeleton (Fig. 1C) was designed to reduce overall envelope. Benchtop tests quantified system-wide closed-loop torque bandwidth, and walking trials quantified torque tracking error while verifying that large torques could be comfortably applied. Both ankle exoskeletons interface with the foot under the heel, the shin below the knee, and the ground beneath the toe. The exoskeleton frames include rotational joints on either side of the ankle, with axes of rotation approximately collinear with that of the human joint (Fig. 1 B, C).

The exoskeletons were designed to provide greater peak torque, peak velocity and range of motion than observed at the ankle during unaided fast walking. The Alpha and Beta devices can deliver peak plantarflexion torques of 120 Nm and 150 Nm respectively. The Alpha and Beta devices both have a range of motion of 30° in plantarflexion to 20° in dorsiflexion.

We tested closed-loop torque control by commanding 50 Nm and 20 Nm linear chirps in desired torque while the exoskeletons were worn by human users.

RESULTS

The total mass of the Alpha and Beta exoskeletons were 0.835 and 0.875 kg, respectively. The gain-limited closed-loop torque bandwidths of the Alpha device with 20 Nm and 50 Nm peak torques were 21.1 Hz and 16.7 Hz, respectively. The phase-limited bandwidths for the Beta device, at a 30° phase margin, with 20 Nm and 50 Nm peak torques were 24.2 Hz and 17.7 Hz, respectively. In walking trials with the Alpha device, the peak average measured torque was 80 Nm. The maximum observed torque was 119 Nm. The root mean square (RMS) error for the entire trial was 1.7 ± 0.6 Nm, or 2.1% of peak torque, and the RMS error of the average stride was 0.2 Nm, or 0.3% of peak torque. With the Beta device, the peak average measured torque during walking was 87 Nm. The maximum observed torque was 121 Nm. The RMS error for the entire trial was 2.0 ± 0.5 Nm, or 2.4% of peak torque, and the RMS error of the average stride was 0.3 Nm, or 0.4% of peak torque.

DISCUSSION

The measured bandwidth, peak torque capability, and mass of both exoskeletons compare favorably with existing devices and with human ankle musculature. In walking tests, we demonstrated robust, accurate torque tracking and the ability to transfer large dynamic loads comfortably to a variety of users. Our results demonstrate that these exoskeletons can be used to rapidly explore a wide range of control techniques and robotic assistance paradigms as the primary elements of versatile, high-performance testbeds.

The three point interface featured in both devices was comfortable and effective. Leaf springs and coil springs were effective series elastic elements. Leaf springs were lighter, but also less robust and more bulky than the coil spring. Strain gauges used in the Beta device were lighter, less expensive, and more effective than the load cell used in the Alpha exoskeleton. The reduced envelope of the Beta exoskeleton was achieved by an acceptable reduction in compliance. These results provide insights into desirable properties of lower-limb exoskeleton hardware, which may inform future designs.

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