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Design of a Low Profile, Unpowered Ankle Exoskeleton That Fits Under Clothes: Overcoming Practical Barriers to Widespread Societal Adoption

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Abstract

Here we present the design of a novel unpowered ankle exoskeleton that is low profile, lightweight, quiet, low cost to manufacture, intrinsically adapts to different walking speeds, and does not restrict non-sagittal joint motion; while still providing assistive ankle torque that can reduce demands on the biological calf musculature. This work is an extension of the previously-successful ankle exoskeleton concept by Collins, Wiggin, and Sawicki. We created a device that blends the torque assistance of the prior exoskeleton with the form-factor benefits of clothing. Our design integrates a low profile under-the-foot clutch and a soft conformal shank interface, coupled by an ankle assistance spring that operates in parallel with the user's calf muscles. We fabricated and characterized technical performance of a prototype through benchtop testing and then validated device functionality in two gait analysis case studies. To our knowledge, this is the first ankle plantarflexion assistance exoskeleton that could be feasibly worn under typical daily clothing, without restricting ankle motion, and without components protruding substantially from the shoe, leg, waist or back. Our new design highlights the potential for performance-enhancing exoskeletons that are inexpensive, unobtrusive, and can be used on a wide scale to benefit a broad range of individuals throughout society, such as the elderly, individuals with impaired plantarflexor muscle strength, or recreational users. In summary, this work demonstrates how an unpowered ankle exoskeleton could be redesigned to more seamlessly integrate into daily life, while still providing performance benefits for common locomotion tasks.

Keywords

human augmentation; wearable assistive device; exosuit; ankle-foot orthosis; ankle plantarflexors

I. INTRODUCTION

A VARIETY of wearable ankle exoskeletons have been developed to augment human locomotion, to enhance healthy gait or assist individuals with musculoskeletal or neurological injuries [1]. However, only recently have portable, untethered exoskeletons demonstrated an ability to reduce the metabolic cost of gait for healthy individuals [1]–[4] (Fig.1). Improving walking economy in healthy individuals is a deceptively difficult task, due to human walking expertise[2], [5] and adoption of near-optimal gait patterns [2], [6]–[11]. Meanwhile, minor or unintended perturbations to normal gait, as well as added weight due to devices, incur metabolic penalties and affect natural kinematics [2], [7],[12]–[15]. Exoskeletons, whether designed to assist healthy or impaired populations, must therefore be thoughtfully designed to interface with the human body and carefully controlled to provide appropriately-timed assistance, without impeding gait dynamics [15], [16].

Recent powered (active motor-controlled) and unpowered (passive spring) exoskeletons have demonstrated the ability to reduce metabolic cost [2]–[4] and muscle activity [2] during walking by assisting ankle moment (i.e., torque) and power provided by the plantarflexors. Throughout much of the stance phase in walking, the calf muscles contract nearly isometrically [17]–[22], supporting a considerable ankle moment [2],[17]. During this time, the Achilles tendon stretches, storing energy elastically [17], [19]. At the end of stance, termed Push-off, the calf muscles contract concentrically, generating positive work [2], [17], [19], [20]. This muscle work, in combination with elastic energy return due to tendon recoil, contributes to accelerating the swing leg and body's center-of mass [23]. Ankle exoskeletons typically act in parallel with the calf muscles, offloading forces on biological muscles and/or reducing power generated through muscle fascicle contraction. Since muscle contractions consume metabolic energy, reducing muscle demands (i.e., force or power) has the potential benefit of also reducing the biochemical energy needed to fuel movement.

A key characteristic that differentiates recent exoskeletons from traditional ankle-foot orthoses (AFOs) is the ability to assist during stance phase and to get out of the way during leg swing [24]. During both stance and swing phases of healthy gait the ankle dorsiflexes beyond neutral (foot orthogonal to shank). If a traditional AFO were designed with a high stiffness (which provides a moment when the ankle is beyond a neutral angle) in order to assist the ankle plantarflexors during stance, then it would interfere with (i.e., resist against) the ankle dorsiflexors during leg swing, impeding one's ability to lift their foot up (which helps reduce trip risk) [31], [32]. Restricting ankle motion can also limit a person's ability to perform various common activities such as running and stair descent, decreasing locomotion performance [1], [24], [33]. If an AFO were designed with low stiffness about the ankle, then it may be too weak to augment the ankle plantarflexors, given the substantial moment these muscles support during stance phase. The most common clinical AFOs are designed with low stiffness about the ankle in order to assist the dorsiflexor muscles (not plantarflexors); specifically to avoid foot drop during leg swing due to neurological or musculoskeletal disorders [31], [32]. Other AFO design variants also exist which can independently adjust plantarflexion and dorsiflexion stiffness, or spring set point (e.g., NeuroSwing, Fior and Gentz), but each one still has to trade-off device performance during stance (i.e., magnitude of elastic energy storage, and portion of gait cycle over which AFO is

assisting) against freedom of movement during swing (i.e., ability to dorsiflex or plantarflex the ankle without interference from the AFO). For instance, an AFO with a neutral or slightly dorsiflexed set point would reduce interference in swing; however, at the expense of reducing elastic energy storage during stance (relative to energy storage with an earlier, i.e., more plantarflexed, set point).

Exoskeletons have overcome this critical limitation of AFOs through a bimodal (or multi-modal) behavior that enables them to assist the ankle plantarflexors in stance without impeding the dorsiflexors in swing. In powered exoskeletons this is accomplished through control or back-drivable actuator-transmission systems, and in unpowered exoskeletons this behavior is achieved by clutching/unclutching a stiff plantarflexor assistance spring so that it does not interfere with swing phase ankle kinematics [2], [34]–[38]. Note that for simplicity throughout this manuscript we use the term exoskeleton colloquially to refer to the general category of wearable devices that physically augment human movement, including soft exosuits and clothing-like assistive devices.

For exoskeletons to be practical in daily life – particularly for individuals who are healthy, or those who experience minor pain, fatigue or disability – assistive devices may need to be lightweight and form-fitting so that they can be worn without interfering with everyday activities, natural movement patterns or physical appearance. Existing ankle exoskeleton designs (Fig. 1) contain elements (e.g., motors, gears, battery packs, lever arms) that protrude out from the foot, leg, waist or back. These protruding elements typically prevent these devices from being worn inconspicuously under normal clothing, which is often preferred by users who want to wear devices in public settings [39], [40]. Also, some mechanisms (e.g., motors, metal clutches) are loud, drawing unwanted attention to the user. One concern is that the inability to conceal devices (visually, audibly, etc.) may discourage widespread adoption of exoskeletons in society, reducing the number of people receiving health benefits from these technologies. Protruding elements can also interfere with daily activities, such as stair descent (lever arm may hit stair) or driving a car (lever arm may hit floor or seat), and may in some cases pose a safety risk (e.g., protruding features may cause tripping hazard or get caught on surrounding environments, such as in a hospital or factory). Compounding these form-factor concerns is the potential issue of artificial joints in exoskeletons, which can affect user performance or comfort if misaligned [14], [41]–[44]. Most (but not all) previous ankle exoskeletons have employed a hinge-like artificial joint between the shank and foot. This can introduce misalignment issues between the artificial joint and biological joint (which is not a perfect hinge), and also restrict non-sagittal plane ankle motion, which may inadvertently affect stability or agility. Moving forward, a key challenge in the exoskeleton field is to achieve functional performance benefits that are similar to (or even better than) existing devices, but with new or refined designs that reduce the device form-factor and conspicuousness, and minimize interference with activities of daily living.

Here we sought to iterate upon a previously-successful unpowered ankle exoskeleton developed by Collins, Wiggin, and Sawicki [2]. The basic functionality of this ankle exoskeleton is that a stiff assistance spring acts in parallel with the biological ankle plantarflexor muscles from early through late stance (offloading the calf muscles), and then

this spring passively disengages during leg swing to allow the person to dorsiflex their ankle naturally without interfering. To accomplish this, the physical structure of the previous device (Fig. 1C) utilizes a clutched spring acting about an artificial ankle joint, joined via a carbon fiber molded frame. We aimed to retain this core functionality that augments walking while reducing the form-factor to fit underneath clothing, removing rigid components that restrict non-sagittal ankle motion (e.g., inversion/eversion), and reducing noise due to the ratchet clutch mechanism. Effectively, we sought to create a passive, clothing-like device that could inconspicuously assist the ankle in daily life. In order to achieve this objective we invented a new type of low profile, under-the-foot clutch mechanism and physically attached the device to the user's shank using a soft, conformable interface. The purpose of this manuscript is to detail these design features, summarize how we integrated them into a physical prototype, then characterized technical performance of the prototype and used human subject case studies to demonstrate feasibility of this lightweight, quiet, low profile, and form-fitting exoskeleton to augment ankle dynamics.

II. DESIGN OVERVIEW

A. Summary

Key aspects of our design include (i) a novel, low profile, under-the-foot clutch mechanism, and (ii) a soft, conformal shank interface sleeve that removes the need to include an artificial ankle joint (e.g., hinge). These two elements of the design (detailed below) are then coupled together by a stiff extension spring, termed the assistance spring (Fig. 2). The shank interface is positioned just below the knee and attached to the assistance spring, which essentially serves as an artificial Achilles tendon. The assistance spring connects to a fibrous rope, which runs posteriorly along the shank down to a lever arm at the heel. From the heel the rope is routed through a rope guide and attaches to the low profile clutch mechanism located under the foot.

B. Overview of Device Function

Our device was designed to mimic the core functionality of the prior exoskeleton during walking [2]. During stance, the clutch is engaged and the stiff assistance spring stretches (Figs. 3–4). A portion of the ankle moment is borne through this spring, and thus reduces force demands on the biological calf muscles. During swing, the clutch disengages (Figs. 3–4) to allow the user to dorsiflex their ankle normally, encountering only small resistance due to a weak reset spring and slight friction within the clutch mechanism. This spring engagement/disengagement behavior is achieved passively (Fig. 4), and thus our device function is qualitatively similar to [2]. Therefore, the major contribution of this work is achieving similar functionality with a lower profile, quieter, and less restrictive design.

C. Clutch Mechanism

We invented a new, unpowered, friction clutch mechanism to engage the stiff assistance spring during the stance phase of gait, and disengage this spring during leg swing (Figs. 3–4). Our clutch design is conceptually similar to a prosthetic ankle mechanism developed by Nickel et al. [45]; however, our physical implementation of the mechanism is quite different. What makes our design unique is that we moved the clutch underneath the foot. The clutch

uses friction due to body weight for the clutching function, enabling it to operate across various gait speeds (as demonstrated in one of the case studies, detailed below). The clutch can be located either above the insole, below the outsole, or potentially integrated into the shoe itself between the insole and outsole.

The friction clutch consists of 5 core elements: reset spring, slider, top gripper, bottom gripper, and spacer (Fig. 4A). In the current prototype, the reset spring is a thin, flat elastic sheet of natural rubber latex (0.25 mm thick, 50 mm wide and 70 mm long, 1.3 kN/m stiffness) located at the distal end of the foot (near the toes). This spring is analogous to the tensioning spring detailed in [2], and is intended to reset the system in early stance (see Figs. 3–4) and to provide minimal resistance to ankle motion during swing (Figs. 5–7). In theory, the reset spring stiffness could be infinitesimally small, but in practice it has to be able to overcome any friction in the system (e.g., in rope guide or the slider rubbing against the gripper). Friction depends on many design and fabrication factors and is difficult to predict analytically. Therefore we tested several stiffness values (see Section III for methods and quantitative results) and used these empirical data to select a reset spring which had sufficient stiffness to reset the system (i.e., overcome friction) and was robust enough for extended human subject testing.

The slider is comprised of a thin composite layer of grip tape (0.5 mm thick) and Nylon canvas fabric (0.5 mm thick), which is connected in series with the reset spring. The slider is located in between the top and bottom gripper layers. Gripper layers are made of rubber (each layer 1 mm polyvinyl chloride). Both the slider and the grippers are flexible (i.e., bendable) so as to not interfere with the natural bending motion of the foot or shoe. A thin Dyneema rope (1.8 mm diameter) is attached in series to the proximal end of the slider, which is then routed through a gap in the Kydex heel and through a low-friction plastic tube that follows the natural contour of the heel (Fig. 4) before connecting up to the assistance spring and then to the shank interface sleeve.

Kydex thermoplastic is used as a thin spacer between the gripper layers. When body weight is not on the foot, then the spacer allows enough room for the slider to move freely between the grippers. The spacer is thin (3 mm tall) such that when body weight is applied, the gripper layers squeeze together to restrict motion of the slider. The spacer height used was determined through iterative testing and empirical observation. Other spacer heights could also be chosen, but the design trade-off is that if the spacer height is too large then the mechanism may not clutch reliably, and if the height is too small then increased friction during swing phase would require a stiffer reset spring. The spacer is designed with a solid heel (Fig. 4A) to prevent the grippers from clutching the slider at heel contact and before foot flat. Gaps in the Kydex spacer near the metatarsals allow for natural toe break (i.e., toe joint articulation [46]).

A metal lever arm (low-carbon steel) extends 36 mm behind the heel of the shoe, in order to amplify assistive ankle torque, while still allowing our device to fit comfortably under common clothing. We found this to be a practical lever arm length during the design iteration process (a trade-off between providing a larger mechanical advantage about the

ankle vs. fitting easily under clothing), but this length could be further optimized for specific tasks, or personalized for specific users.

D. Shank Interface

We developed a lightweight, conforming sleeve-like interface to attach the exoskeleton to the shank of the user. The inner-most layer of the sleeve is comprised of a 5 mm thermoplastic elastomer gel liner material covered with a hook- and-loop compatible cloth (Willowood Alpha Classic liner with Spirit Fabric). This type of liner material is commonly used in the field of prosthetics, as it adheres well to a person's limb, is skin-friendly, and can undergo considerable shear loading without slipping down the limb. A semi-rigid plastic (or alternatively, nylon fabric) outer shell was then custom- molded and attached to the liner via hook-and-loop. This outer shell distributes forces across the surface area of the liner and provides an anchor point to attach the assistance spring. A hook-and-loop strap was attached around the shank at the top of the interface and used to provide light compression and distribute forces. Our interface design is conceptually similar to previous soft interfaces used in powered soft exoskeletons (i.e., exosuits [13], [14], [28], [47]–[49]), though using different materials. This style of interface eliminates the need for an artificial (exoskeleton) ankle joint, which is different from the majority of existing ankle exoskeletons [2], [25], [26], [50],[51]. Collectively, these features result in a low profile shank attachment that can be worn under normal clothing while sustaining the needed loads for ankle torque assistance.

E. Prototype Specifications

The total weight of our prototype was 459 g, of which 263 g was due to the shank interface, 10 g was due to the assistance spring, and 186 g was due to the clutch. Our current prototype weighs substantially less than previously developed powered ankle exoskeletons (~1000–1500 g per leg plus 2500–5000 g attached at the waist/back, Table I). Our prototype is comparable to (or even slightly lighter than) the weight of many traditional plantarflexor assist AFOs (Table I). Our prototype is also comparable to the weight of the unpowered exoskeleton developed by Collins, Wiggin, and Sawicki ([2], Table I). Note that dropfoot AFOs are lighter than all other AFOs and exoskeletons because they only need to provide a small torque to support weight of the foot during leg swing (Table I). By using a thinner gel liner (e.g., with 3 mm thick thermoplastic elastomer), we expect that the weight of our device could be further reduced. The total height of our prototype under the plantar surface of the foot/shoe is 5.1 mm (i.e., comparable to, or even shorter than, the thickness of a typical shoe insole), and the device extends less than 40 mm from the shank (though this could be even less depending on choice of lever arm length). The effective lever arm length is about 100 mm, measured as the perpendicular distance from the ankle joint center to the assistance spring line of action (and thus is somewhat dependent on user anatomy). The assistance springs used were metal extension springs of roughly linear stiffness. However, this spring can be interchanged with other elastic elements (e.g., springs, bands, cables) to individualize or optimize assistance. Indeed, as detailed in our case studies below, we swapped in and tested multiple different springs (6.1–35.4 kN/m), which encompassed the stiffness range previously tested [2]. Note that elastic elements could be of linear or non-linear stiffness. For instance, a simple stiffening spring could be achieved by placing two (or

more) elastic elements in parallel, and then setting one (or more) of the spring rest lengths to be longer than the other(s). Our prototype contains approximately \$100 in materials of which \$67 was for the prosthetic liner material. All components were purchased in low volume. We therefore expect that our exoskeleton would be inexpensive to manufacture in bulk.

III. DEVICE CHARACTERIZATION

A. Methods

Our current prototype was developed through an iterative design process. We then built versions of this prototype and performed quantitative benchtop tests to characterize/benchmark technical performance. First, we characterized the maximum holding force of the clutch and estimated the clutch friction coefficient. We applied downward (i.e., normal) forces to the clutch using metal weights (to simulate the vertical ground reaction force, Fig. 5A), ranging from 100–900 N. At each weight we used a computer-controlled robotic actuator (Humotech, Pittsburgh, USA) to apply pulling forces to the fibrous rope connected to the clutch, while a load cell in series recorded forces. The actuator provided ramped pulling forces, which progressively increased until the clutch slipped (observed visually, and confirmed via load cell). The peak force achieved before slippage occurred was defined as the maximum clutch holding force for each normal force magnitude. The average time to the peak force was similar to walking at 1.25 m/s (~0.5 seconds, see Fig. 6). We performed this characterization test twice: once while the clutch was set flat on the ground (to simulate early and mid-stance) and next while the heel was lifted off the ground at ~60° (to simulate Push-off). For each configuration (foot-flat and heel-lift), the maximum clutch holding force was plotted as a function of normal force, a linear regression was fit, and the slope was computed to estimate the average coefficient of friction. We then combined these estimated friction coefficients with published data on vertical ground reaction forces during human walking [52] in order to compute a theoretical maximum amount of assistive ankle torque that could be provided by our current prototype without clutch slippage (Fig. 5B).

Next, we quantitatively characterized the reset spring. We built three versions of the clutch, each with different reset spring stiffnesses (0.7, 1.3 and 1.9 kN/m). We then had a subject don the full prototype (Fig. 2), lift their foot into the air and rotate their ankle through a full range of plantarflexion/dorsiflexion motion. We estimated ankle rotation and the torque due to the reset spring (using measurement methods detailed in section IV-A), then plotted ankle angle vs. reset spring torque (Nm/kg) for each of the stiffness values (Fig. 5 Inset).

B. Results

We found the average friction coefficient in the flat configuration to be 0.79, and in the Push-off (heel-lift) configuration to be 0.58 (Fig. 5A). Applying these friction coefficients to healthy walking data from [52] at 1.25 m/s we estimate that the maximum exoskeleton ankle torque that could theoretically be provided by our current prototype would be 0.75 Nm/kg at this speed (Fig. 5B). We also note that at the point in the stride where peak ankle dorsiflexion angle occurs (~50%, at the discontinuity in Fig. 5B), this max theoretical exoskeleton torque was ~0.6 Nm/kg. These magnitudes are 100% and 60% larger,

respectively, than peak torques reported in the experimental study by Collins, Wiggan, and Sawicki [2]. Peak exoskeleton torques over the full ankle range of motion (due to the reset spring and friction in the system when the foot was unloaded) were less than 0.008, 0.011, and 0.018 Nm/kg, for the 0.7 kN/m, 1.3 kN/m and 1.9 kN/m reset springs, respectively (Fig. 5 Inset). We also confirmed from video recordings that each of the three springs tested was able to overcome friction in the system and therefore fully reset the slider (i.e., return it back to its default position).

IV. CASE STUDY I – VARYING STIFFNESS, EFFECT ON EMG

A. Methods

All human subject studies were approved by the Vanderbilt University Institutional Review Board and the subjects gave informed written consent prior to participation. The first case study was performed to demonstrate that our device functions on a real user, and that it can provide calf muscle offloading (similar to [2]). One male subject (Age: 22 years old, Height: 1.7 m, Mass: 84 kg) walked on a level, split-belt force- instrumented treadmill at 1.25 m/s. Prior to testing, the subject walked with the exoskeleton prototype for over an hour to acclimate to it. The subject then returned for a separate gait analysis session. The subject walked normally in casual shoes (Shod condition, Etnies Fader LS) and then with the exoskeleton added onto the same shoes and worn unilaterally. We tested 3 different assistance springs (6.1, 13.2, and 17.7 kN/m, chosen to be a similar range to [2]). The subject walked with each spring for two minutes during testing. We collected ground reaction forces (Bertec, 2000 Hz), electromyography (EMG, Delsys, 2000 Hz) of the soleus muscle, and lower limb kinematics (Vicon, 200 Hz). A heel raise trial at the beginning of the experiment was used to estimate EMG during maximum voluntary contraction (MVC).

Ground reaction forces and motion data were bi-directionally low-pass filtered by a 3rd order Butterworth filter with a cutoff frequency of 20 Hz. EMG data were demeaned, high-pass filtered (150 Hz cutoff), full wave rectified, and low-pass filtered (10 Hz cutoff), similar to [53], [54]. The resulting EMG envelopes were then normalized by the maximum activation among the trials (walking and heel raise MVC) [54]. Ankle exoskeleton torque was estimated using spring constants, measured spring displacement from motion capture, and the exoskeleton lever arm (assumed constant). Exoskeleton torque was normalized by subject body mass. Data were then split into strides (foot contact to ipsilateral foot contact) using ground reaction force data. Average (mean) EMG activity was computed per stride as the time integral of the EMG envelope divided by the stride time, similar to [2]. To ensure steady-state had been reached, we averaged across the final 10 strides from each trial and then plotted comparisons between conditions. We expected that soleus muscle activity would be reduced when walking with the exoskeleton relative to the shod condition, similar to [2]. Finally, to assess quietness of our device we had the subject walk over ground and used a decibel (dB) meter to quantify peak noise levels shod vs. while wearing our ankle exoskeleton prototype.

B. Results

We confirmed prototype function using high-speed video of the subject walking (i.e., we observed evidence of clutching and stretching of the assistance spring during stance, followed by unclutching and stretching of the reset spring during leg swing). Over the stride, we observed reductions of 5–17% in average soleus activity for the exoskeleton conditions relative to the Shod condition (Fig. 6). Reductions in soleus EMG were most evident throughout the middle of stance phase. Peak exoskeleton torque with the stiffest spring was about 0.2 Nm/kg during stance. We observed dorsiflexion torque during swing of <0.04 Nm/kg (Fig. 6). EMG reductions and the shape of the exoskeleton torque curve were qualitatively consistent with findings from [2]. In terms of audible noise, walking with the exoskeleton was similar to walking in normal shoes (62 dB peak with the exoskeleton vs. 60 dB peak in normal shoes, measured in a room with 57 dB ambient noise).

V. CASE STUDY II – VARYING SPEED, DOUBLING STIFFNESS

A. Methods

The second case study was intended to demonstrate that the prototype works over a range of walking speeds, and to explore the effects of stiffer assistance springs on exoskeleton ankle torque. A different male subject (Age: 27 years old, Height: 1.8 m, Mass: 93 kg) was recruited. Prior to testing, the subject walked with the exoskeleton prototype for over an hour to acclimate to it. The subject then returned for a separate gait analysis session. The subject first wore the exoskeleton unilaterally (spring stiffness of 17.7 kN/m) and walked at speeds varying from 0.5–1.75 m/s on a level, split-belt force-instrumented treadmill while we collected ground reaction forces and lower limb kinematics. Next, we doubled the assistance spring stiffness (to 35.4 kN/m), and had the subject again walk over this range of speeds. Using this stiffer spring, we also had the subject adopt a static tandem stance posture with the exoskeleton on the trailing limb, then lean forward to maximally dorsiflex their ankle, which enabled us to estimate the highest exoskeleton torque achievable with this spring and to confirm that both the under-the-foot clutch and shank interface could provide the necessary holding forces. Data were processed with the same methods described in Case Study I.

B. Results

The exoskeleton functioned over the range of gait speeds, as evidenced by assistance spring motion and torque profiles (Fig. 7). For the 17.7 kN/m spring, peak exoskeleton torque was 0.12 Nm/kg at slow speeds, increasing to a maximum of 0.14 Nm/kg at moderate speeds, then decreased to 0.11 Nm/kg at the fastest speeds (Fig. 7). When spring stiffness was doubled, the exoskeleton provided approximately twice the torque over the same speed variations (Fig. 7). During maximum torque tests, the exoskeleton provided an average maximum 0.62 Nm/kg at a peak dorsiflexion angle of 19 degrees.

VI. DISCUSSION

Here we present a novel design for an unpowered ankle exoskeleton that is low profile, lightweight, quiet, suitable for different gait speeds, low cost to manufacture, and not

restrictive of non-sagittal degrees-of-freedom; while still providing assistive ankle torque that can reduce demands on the biological calf musculature. This work is a direct extension of the previously-successful ankle exoskeleton concept presented by Collins, Wiggin, and Sawicki [2]. Our key contribution is in creating a device with similar function, but within a form-factor that fits more seamlessly into daily life. To our knowledge, this is the first ankle exoskeleton that could be feasibly worn/concealed under typical daily clothing, without restricting non-sagittal ankle motion, without elements protruding from the leg or shoe, and without auxiliary components (e.g., actuators or batteries) carried on the back or waist. Our new design highlights the potential for performance-enhancing exoskeletons that are inexpensive, unobtrusive, and can be used on a wide scale to benefit a broad range of individuals throughout society. Societal applications of our ankle exoskeleton might include: (i) assisting the elderly [55] or individuals with impaired plantarflexor muscle strength (due to neurological or musculoskeletal injuries) to augment their physical capabilities and help keep them active, (ii) assisting recreational users in order to reduce musculoskeletal loading and effort during walking, running or hiking, or (iii) assisting users who walk or run for substantial lengths of time to reduce fatigue and help keep them energized and productive (e.g., postal or warehouse workers, soldiers).

A. Design Innovations

Our design has three key benefits which facilitate integration into daily life. First, our design reduces physical bulk and range-of-motion restrictions. There are no components that protrude substantially from the foot, shank, waist or back, and there is no artificial joint restricting ankle motion. In conjunction with the shank interface, this allows our device to be worn easily underneath typical clothing and in/under shoes, without interfering with other movement degrees-of-freedom (e.g., ankle inversion/eversion) or common daily tasks (e.g., driving a car). Devices of this type could conceivably be integrated under, or even into everyday clothing or uniforms (e.g., a specialized pair of shoe insoles and socks), which is important to some exoskeleton users [39], [40]. Using body weight to clutch a tensile element (the slider) via friction allows us to use thin, lightweight and flexible materials. Lightweight devices have value to clinical populations, and thus this feature promotes future adoption by these users [1], [24]. This also means our device can be easily integrated into either the insole or outsole of a shoe and is not expected to interfere with natural foot kinematics or footwear mechanics/comfort. Anecdotally, this was supported by feedback from case study subjects, who reported no adverse reaction to our prototype over the course of >1 hour of walking with the device.

Second, our design is fully passive yet intrinsically adapts its behavior to suit a wide range of gait speeds without relying on electrical sensors or specially-tuned mechanisms. We observed that our clutch reliably engaged and provided assistive torque across a range of gait speeds (Fig. 7). Prior devices have relied on foot switches (which are not always reliable, [56], [57]) or an adjustable ratchet mechanism (which relies on the ankle going through a sufficient range of motion [2]). The advantage of the under-the-foot clutch is that it is naturally engaged during stance phase and disengaged during leg swing, essentially using the user's body weight as the control signal and alleviating the need for any motors. Thus clutch engagement/disengagement is not an explicit function of time or joint angle, but

rather a function of foot contact with the ground. Being a completely passive device has additional advantages: there is no added weight due to batteries, the device could be used for extended periods of time with zero access to the power grid, and in an increasingly digital world our exoskeleton would be one less thing users have to remember to plug in.

Third, the materials used in the clutch enable our design to be quieter than previously-used metal clutches (e.g., using ratchet mechanisms), and thus audibly more similar to quieter electrostatic clutches [58]. Audible device noise is a substantial societal barrier to the wider adoption of exoskeletons, as many people are embarrassed when loud noises they emit draw unwanted attention (farting is a good example). This is a common human experience, and evident in the feedback we have received from multiple clinical populations. Our device's noise levels are similar to walking in normal shoes (62 vs. 60 dB), and thus could be worn in public settings without undue attention. Our clutch design provides a novel way to retain assistive exoskeleton function, minimize form-factor and overcome this audible noise impediment. Finally, our design is low-cost (~\$100 without economies of scale or optimized design for manufacturing), which is key because device affordability is often important to consumers and end users of exoskeletal devices [59].

B. Feasibility of the New Design

High-speed video observation, benchtop characterization of technical performance, and two instrumented gait analysis case studies collectively provide empirical proof-of-concept of our exoskeleton design. During the case studies, we found that the assistive ankle torque profile vs. time was similar in shape to torque profiles reported in [2]. In Case Study I we observed peak device torques in the range of 0.1–0.2 Nm/kg, which were lower than exoskeleton assistance in the prior study (0.25–0.37 Nm/kg peak, [2]). This was partially because our exoskeleton has a smaller lever arm (100 vs. 150 mm). In Case Study II, we found that doubling the assistance spring stiffness resulted in approximately double the peak exoskeleton torque (>0.2 Nm/kg) for the subject tested. We also found in a static maximum dorsiflexion task that our prototype could support much higher torques (>0.6 Nm/kg, with a 35.4 kN/m spring), consistent with our theoretical model predictions based on empirically-derived coefficients of friction (Fig. 5B). We note that the effective spring stiffness (and therefore assistive torque) is reduced by the compliance of our viscoelastic interface, as evidenced in prior studies [13],[60]. Collins, Wiggin, and Sawicki reported a similar effect in their more rigid device, which had a 33% lower stiffness than the nominal spring value, due to device compliance [2]. Nevertheless, in Case Study I we still observed a 5–17% reduction in soleus muscle activity relative to not wearing the device ($N=1$), which was similar to the 9–14% reported in [2] ($N=9$). Although the EMG reduction in our study exhibited an increasing trend with decreasing spring stiffness, we caution against over-interpreting the significance of this trend given it was a single subject case study. See prior studies [2], [55], [61] for more comprehensive, multi-subject evaluation of how these types of unpowered ankle exoskeletons affect gait kinematics, kinetics, EMG and metabolic cost on average. The key take away is that our new design and current prototype provided ankle torque assistance and reduced calf muscle activity as we expected, while using a different clutch mechanism and a smaller and less rigid form-factor, as compared to the prior exoskeleton [2].

C. Prototype Limitations

The current prototype was developed, built and tested to demonstrate one implementation of our design concept. The prototype would benefit from improvements in materials and manufacturing techniques to improve longevity and robustness of the device, which we expect would also reduce friction in the system when unclutched (swing) and enhance friction when clutched (stance). The weight of our current prototype was largely a function of the shank interface. The thickness of the shank interface (5 mm) was due to the commercially available prosthetic liner we used, but in the future this interface could be made thinner or perforated to reduce weight (e.g., reducing from 5 mm to 3 mm liner thickness would reduce weight by approximately 85 g). We did not perform mechanical fatigue testing on the current prototype. Such testing would be more appropriate after completion of the design for manufacturability process. Nevertheless, we note that our current prototype did operate for over one hour of training/testing without failing or breaking.

D. Case Study Limitations

The case studies were intended to demonstrate proof-of- concept, and specifically that our device functions as expected to augment ankle mechanics. Prior studies (e.g., [2]) provide more extensive evidence of performance benefits that can be obtained through ankle augmentation. We limited our case studies to level walking. We did not attempt to optimize the assistance spring stiffness. Of note, prior experimental testing[55] and theoretical models [62] each suggest that a constant spring stiffness may be near optimal for a fairly wide range of walking speeds. Due to the shorter lever arm and interface compliance, our peak device torque during steady-state walking was lower than the predicate exoskeleton [2] when using similar stiffness springs. However, we demonstrated that our exoskeleton was capable of producing higher levels of torque with greater assistance spring stiffness (e.g., Fig. 7). During the static maximum torque trial, using the stiffest spring tested in our case studies, the shank interface supported about 650 N of peak downward force. Previous published studies have also successfully applied ~300–500 N to similar shank interfaces without user discomfort [13], [47], [63]. What remains to be seen is if these higher levels of force can be sustained for longer periods of time during walking without subject discomfort (which is important for assistive device users [1],[59]). Alternatively, slight increases in the lever arm would increase assistive ankle torque and reduce applied forces, while likely still being concealable under most clothing. Additional modes of locomotion (e.g., running) and terrains (e.g., slopes) warrant formal multi-subject testing in future studies, as does evaluation of non-linear vs. linear springs.

We did not test users with alternative gait patterns (e.g., equinus gait or other patterns observed in clinical populations), which could affect the performance of the device. Our exoskeleton is intended for users who can achieve sufficient ankle dorsiflexion, and who may benefit from low to moderate torque assistance. For populations which require more substantial amounts of assistance or more specialized/variable control, then a powered device, rigid frame or alternative intervention may be required. In these cases, powered assistance and/or rigid frame components may be valued more highly than device size, weight or concealability.

E. Translational Challenges

Our new device design addresses some of the key challenges to translating assistive devices from the lab into the real world. However, we foresee a few additional potential challenges related to exoskeleton translation: interface migration, effects on thermoregulation and sweat. We did not observe migration of the shank interface (i.e., slippage relative to the skin) during our case study experiment; however, it has been observed that these types of sleeves can sometimes begin to slowly migrate down the leg over longer timescales or in the presence of substantial sweat [13]. Prior work suggests that such interface migration can be mitigated, in part or in whole, by further tensioning the strap above the calf muscle bellies [47], [48], [63]. Migration is important to mitigate because it can impact both device performance and user comfort. Differences in the shape of each user's shank also affect device fit (and potential migration) and should be considered in the design of the interface. In terms of translating our device into real-world environments, we predict that heat and sweat are also pragmatic challenges to address. These issues could potentially be mitigated by various means. For instance, the sleeve might be designed to dynamically loosen/tighten on the shank, through the use of shape memory alloy actuators [64] or a compact manual lacing system (e.g., Boa Lacing System). In effect, this would allow users to switch between a tighter-fitting compression-like garment and looser fitting pants as needed, depending on whether or not they required assistance from the device. Alternatively, sweat might be controlled with locally applied antiperspirants, or thermoregulation of skin via the inclusion of air flaps/vents, breathable/sweat wicking garment materials or other coolant mechanisms. For certain populations (e.g., elderly with frail skin), comfort and/or the magnitude of loading that can be distributed over the skin may also need to be considered. Although comfort was not a limiting factor in our case studies on young healthy individuals, it may nonetheless place practical limits on the magnitude of assistance that can be provided to certain aging or patient populations, or limits on the amount of continuous device usage by a given user. These areas require future investigation on specific populations, and comfort-related aspects may be highly user-specific.

In future design iterations it may also be preferable to add a small, secondary mechanism on the shank interface which would allow a user to completely disengage the assistance spring (i.e., to completely "turn off" exoskeleton assistance). This functionality would provide two key benefits. First, some individuals may only desire intermittent assistance (e.g., when they are becoming fatigued). In many cases exoskeletons are not intended for all-day, everyday use, but rather can provide assistance on demand and as needed to augment users during difficult, onerous or repetitive tasks. This added feature would enable users to engage/disengage assistance on demand without needing to don/doff the exoskeleton. Second, exoskeleton elastic energy storage and return may not be desirable for all tasks. Although our proposed ankle exoskeleton is of appropriate form-factor for most daily activities, there may still be tasks where it would be preferable for exoskeleton assistance to be completely disengaged (even when the foot is in contact with the ground). A potential example is in stair descent, where the ankle primarily absorbs energy, rather than storing and returning energy. A user-controlled on/off feature would be relatively straightforward to add (e.g., a manual clutch or a simple mechanism to slacken the rope) and would prevent the assistance spring

from stretching over the ankle's full range of motion, even if the under-the-foot clutch is engaged.

VII. CONCLUSION

In summary, this work demonstrates how an unpowered ankle exoskeleton could be redesigned to fit underneath clothing and inside/under shoes, such that it more seamlessly integrates into daily life. In effect, our device blends the assistive benefits of exoskeletons with the form-factor benefits of clothing; analogous to our previously developed clothing-like device for offloading the low back [65]. Our ankle exoskeleton design is low profile, lightweight, quiet, low cost to manufacture, intrinsically adapts its behavior to different speeds, and does not restrict non-sagittal ankle range-of-motion. Yet the device still provides assistive ankle torque during gait that can reduce demands on the biological musculature. To accomplish this, we integrated a novel low profile under-the-foot clutch with a soft conformal shank interface. The clutch and interface were then coupled together by an ankle assistance spring that operates in parallel with the user's calf muscles. To our knowledge, this is the first such device that could be feasibly worn while concealed under typical daily clothing, highlighting the exciting potential for wider adoption of exoskeletal technologies into society, which could benefit a broad range of individuals.

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Fig. 1.

Examples of portable exoskeletons that assist ankle plantarflexion. A) Powered exoskeleton from TU Delft uses an electric motor and leaf spring design [25], [26]. The device requires a backpack-mounted controller and battery pack (not depicted). B) Powered exoskeleton from MIT also uses an electric motor plus leaf spring design, but in a different configuration [3], [4]. A battery pack and controller unit are worn at the waist (not depicted). C) Unpowered exoskeleton from Carnegie Mellon and North Carolina State University uses a clutch and spring [2], [27]. D) Powered soft exoskeleton (exosuit) from Harvard uses motorized Bowden cables to assist ankle plantarflexion (and also dorsiflexion) [28]–[30]. A battery pack, and actuator unit are worn at the waist (not depicted). Several ankle exoskeletons have demonstrated objective benefits in terms of reducing metabolic cost or muscle activity during walking. However, existing designs tend to have mechanical elements that protrude out from the leg, foot, back or waist, which could impede widespread adoption of exoskeletal technologies in society; particularly amongst individuals who prefer to wear devices inconspicuously under their normal clothing.

Fig. 2.

New ankle exoskeleton design combines form-factor of clothing with the function and assistive benefits of a previously-successful unpowered exoskeleton [2]. This design integrates into the shoe and under clothing such that it could be worn inconspicuously in daily life. An additional hook- and-loop strap was attached at the top of the interface and used to provide light compression and distribute forces. It was not depicted to avoid visually obscuring the semi-rigid plastic component on the shank.

Fig. 3.

Exoskeleton function. During walking, the stiff assistance spring is engaged (clutched) during stance phase to assist the plantarflexors, then this spring is disengaged (unclutched) during leg swing to allow for normal, unrestricted ankle dorsiflexion. At foot contact the ankle is slightly dorsiflexed beyond neutral (foot perpendicular to shank), and the weak reset spring is in a stretched position. After foot contact, the ankle begins to plantarflex until the foot becomes flat on the ground. This motion allows the reset spring to recoil. Recoil of the reset spring then pulls the unstretched assistance spring to its default position. As the person progresses over their foot throughout stance, the center-of-pressure under the foot (i.e., location of net ground reaction force) also progresses forward. The ground reaction force compresses the gripper upon the slider, clutching it. In other words, the ground reaction force increases the friction force (between the slider and gripper), preventing further motion of the slider. Because the slider is now fixed, dorsiflexion of the ankle stretches the stiff assistance spring. Stretching of the assistance spring stores elastic energy, and also offloads the calf muscles and tendons acting in parallel. During late stance, biological ankle motion reverses and the ankle begins to plantarflex. At this time the assistance spring recoils, assisting biological ankle plantarflexion Push-off power. As the ground reaction forces underneath the foot rapidly decrease towards zero (toe-off), frictional forces between the slider and gripper also decrease, unclutching the mechanism. The slider can then translate freely, with minimal resistance due to stretching of the weak reset spring. This allows the person to dorsiflex their ankle normally during leg swing (to provide toe clearance). Upon subsequent foot contact the cycle begins again. White areas indicate approximate time of

clutch engagement (similar in timing to the stance phase), and grayed areas indicate clutch is disengaged (similar in timing to the swing phase). See Figure 4 for further description of clutch design and function.

Fig. 4.

A) Under-the-foot clutch design. The friction clutch consists of 5 core elements: reset spring, slider, top gripper, bottom gripper, and spacer. Specific materials, construction details and design rationale are provided in the main text. B) Under-the-foot clutch function when disengaged. When no normal force is applied to the grippers during leg swing (and early stance), the assistance spring and slider translate as the weak reset spring stretches/shortens. C) Under-the-foot clutch function when engaged. When normal force is applied due to body weight during stance, the assistance spring can stretch/shorten, while the slider and reset spring are clutched (i.e., unable to move) due to the friction force between the slider and the grippers. See Fig. 3 for further description of clutch function.

Fig. 5.

Clutch characterization results. A) Friction force (F_f , i.e., max clutch holding force) vs. normal force (F_N), for both flat foot (teal) and heel-off (dark blue) configurations. Friction coefficient, μ , for each configuration was estimated via linear regression. B) Theoretical maximum torque (right vertical axis, teal/dark blue curve) that the exoskeleton could provide based on empirical coefficients of friction and typical ground reaction forces (left vertical axis, gray curve) during walking at 1.25 m/s (data from [52]). Inset: Exoskeleton ankle torque (in Nm/kg) for three different reset spring stiffnesses (in kN/m) when foot is

unloaded (i.e., swing phase) and rotated through a full range of ankle plantarflexion (–) and dorsiflexion (+).

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Fig. 6.

Case study results. A) Soleus EMG was reduced, specifically in mid-stance, when wearing the exoskeleton with various assistance springs vs. not wearing the exoskeleton (normal shoes only, Shod). EMG was normalized to maximum muscle activation and averaged across strides. B) Average soleus EMG over stride (dimensionless) was reduced by 5–17% when wearing the exoskeleton. EMG mean and standard deviation are depicted. C) Exoskeleton torque increased as spring stiffness increased. Spring stiffnesses were selected to match range of springs used in the prior study by Collins, Wiggin and Sawicki [2]. Due to larger lever arm the prior exoskeleton provided more ankle torque (shaded gray) for same spring stiffness range.

Fig. 7.

Representative example of ankle exoskeleton functioning across a range of speeds. Results from two separate trials are overlaid. When assistance spring stiffness was doubled from 17.7 kN/m (teal) to 35.4 kN/m (dark blue), the peak exoskeleton torque approximately doubled at each speed.

TABLE I

MASS OF RECENT EXOSKELETONS AND REPRESENTATIVE AFOs

	Recent Exoskeletons				Plantarflexor Assist AFOs		Dropfoot AFOs
	Mejineke et al. 2014	Mooney, Rouse & Herr 2014b	Bae et al. 2018	Collins, Wlugin & Sawicki 2015	Yandell, Tacca & Zelik 2018	Dynamic Brace (Dynamic Bracing Solutions)	
Total Mass (2 legs, waist & back)	8200 g	4660 g	3800 g (unilateral)	816–1006g	918g	1196 g	240 g
Mass per Leg	1500 g	1060 g	not specified	408–503 g	459 g	598 g	120 g
Mass at waist/back	5200 g	2540 g	not specified	0g	0g	0g	0g