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Improved Mobility with a Neutral, Motion-Amplifying Controller for an Experimental Exoskeleton

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ABSTRACT

The number of seniors is rising worldwide. Exoskeleton devices can help seniors regain their lost power, balance, and agility, thus improving their quality of life. Exoskeleton devices and control strategies assist human gait. A common strategy is to use oscillator-based controllers, which "lock in" with the gait and help the subject walk faster using a phase lead characteristic. Such strategies are limited to gait assist only and are less effective in more general movements. These controllers can be detrimental in critical cases such as when the leg needs to execute a fast reactive stepping to stop a fall. We present a control strategy for a hip exoskeleton, which assists human leg motion by providing motion amplification at the hip joint. The controller is "neutral" because it assists any leg motion, not only a gait, and can help avoid falls by assisting reactive stepping. Our control strategy modifies the joint dynamics of the coupled human-exoskeleton system such that the desired dynamic response is achieved while guaranteeing stability. We define assistance as reducing the impedance and increasing the admittance of the coupled system. The dynamic response of the leg is defined by the frequency response profile of the magnitude of integral admittance (torque-to-angle relationship) of the coupled human-exoskeleton system, and assistance occurs when this profile is higher than that of the unassisted leg for all frequencies of interest. Our controller produces hip joint motion amplification and results in larger and faster leg swing motions, and can help recover the seniors' power and agility.

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INTRODUCTION

World Health Organization (WHO) studies have estimated that between 2000 and 2050, the proportion of the world's population over 60 years will double from about 11% to 22% [1]. With aging, the human body tends to lose its power and agility and this drastically affects the range and quality of mobility that we otherwise take for granted. Exoskeleton devices can potentially help aging seniors regain their lost power and agility and make them more independent and self-reliant by helping them perform their activities of daily living (ADL).

Inherently correlated with the loss of power and agility, aging seniors also experience a general degradation of balance capability and an increased occurrence of fall. In the U.S. alone, an estimated 4.05 million seniors were treated in hospital emergency departments for fall-related fractures between 2001 and 2008 [2]. According to WHO's fact sheet on falls [3], adults older than 65 years (seniors) suffer the greatest number of fatal falls, and falls are the second leading cause of accidental or unintentional injury deaths worldwide. Assistive exoskeleton devices [4, 5] can help reinforce the balance capabilities of aging seniors, reduce their chance of fall and of catastrophic injuries, and thereby promote a lifestyle with safe mobility. Mobility is known to have a strong positive effect not only on the physiological processes such as respiratory and circulatory systems, but also on the psychological state and morale of the person.

A number of exoskeleton devices and control strategies have been developed that aim at assisting human mobility. A common strategy is to use controllers that are based on oscillators [6, 7] such as van der Pol [8] or Duffing [9] or perhaps employ central pattern generator (CPG) [10]. Such controllers lock in with the gait cycle and through phase lead creates the effect of forward push to the joint motion. However, such strategies are limited to assisting cyclic motions and are mainly applicable to gait assist. For general movements and activities that involve frequent variations including start/stop and direction change, these controllers are less effective. In fact, these controllers can be detrimental in critical cases such as when the leg needs to execute fast reactive stepping in order to stop a potential fall.

It has been conjectured that the ideal joint level behavior of an assistive exoskeleton should be that of a pure torque source. In other words, the physical and dynamics footprint of the exoskeleton on the subject should be as small as possible, and should be manifested only through the assistive torque. This translates to the fact that the device, ideally, should be able to perfectly compensate for its own dynamics. Additionally, the exoskeleton should *never* oppose the human. This means that the exoskeleton's torque profile should be in harmony with the motion intent of the person. The other desirable exoskeleton properties are that its controller should be highly responsive (high bandwidth) and stable, and the device should be minimally invasive

with sensing and actuation. Finally, and quite importantly, the exoskeleton should not exhibit unrelated detrimental effects e.g., a gait assist exoskeleton should not negatively affect balance capability.

We propose that exoskeletons that are meant for a general range of activities must additionally possess the property of *neutrality*. A neutral exoskeleton is one that can assist the human joint at a fundamental "joint motion" level and is not partial to a specific motion pattern or a specific activity. Implied in this description is the fact that the exoskeleton controller does not control any "higher level" description of the motion undertaken by the human. Consequently, it simply assists the motion initiated by the human and does not modify the motion with any higher purpose. The philosophy behind this controller is that the human possesses the knowledge of the best limb movement that is required in any given situation, even though they may not always have the physical ability to execute it promptly or fully. The intended effect is that the control law executed by the coupled human-exoskeleton system is unchanged from what the human itself wanted to execute, only that the exoskeleton makes available additional joint torque.

In this paper, we present a novel control strategy for the Honda Stride Management Assist (SMA) exoskeleton device [11]. The SMA device is a 2-DOF exoskeleton device meant for assisting the hip joints as shown in Figure 1. Our control approach, called the *Integral Admittance Shaping*, assists the human by virtually modifying the human leg dynamics such that its effective impedance is reduced and effective admittance is increased. This provides motion amplification (equivalently, torque reduction) to the hip joint. The controller is neutral in the sense described above. In addition to assisting in walking, this controller is equally suited for helping humans in more general ADL activities. Particularly beneficial is its potential role in fall avoidance by facilitating a faster and longer step in reactive situations.

Figure 1. Honda Stride Management Assist (SMA) device.

In the present implementation of integral admittance shaping, the exoskeleton exhibits active impedance at its interaction point with the human body (the location of fastening straps just above the knee joint). In simplest terms, active impedance can be thought of as the generalized version of a negative mass, though the effects of a negative spring and negative damper are also present in the impedance. A negative mass is an active element and the control system displaying an active property is unstable. Our controller achieves active impedance through the *positive* feedback of position, velocity and acceleration. Consequently, this controller is unstable in isolation. One may legitimately ask the wisdom behind deliberately creating an unstable exoskeleton controller. The response to this concern lies in the concept of coupled stability.

The human body, by virtue of its inherent visco-elastic properties, is capable of stabilizing interacting unstable systems through contact. We have experienced this when, for example, by simply touching a vibrating machine component, we can render it stable. While our isolated exoskeleton, displaying active impedance, is unstable, the coupled human-exoskeleton system can still be stable. In this work, we show how active exoskeleton controllers can be designed while still guaranteeing coupled stability.

While the coupled human-exoskeleton system must satisfy the coupled stability condition, we take into account an additional level of system interaction, in which the coupled human-exoskeleton system interacts with a third entity, such as the ground. If the coupled human-exoskeleton system possesses the property of passivity [12], it remains stable while interacting with other passive systems. Noting that the ground and other typical environments are passive, we require that our integral admittance shaping controller guarantees both coupled stability and passivity.

In this paper, the controller is implemented as an optimization algorithm that determines the effective moment of inertia, joint damping coefficient and joint stiffness coefficient for a desired assisted human limb dynamics. We experimentally demonstrate that our controller produces hip joint motion amplification and results in larger and faster leg swing motions on humans. Thus, our controller has the potential to recover the lost power and agility of the lower limbs to aging seniors.

EQUATIONS OF MOTION OF A COUPLED HUMAN-EXOSKELETON SYSTEM

In this work, we use 1-DOF linear models for the leg and the exoskeleton. The 1-DOF leg model is an approximation of the extended leg with locked-knee swinging about the hip joint. This model captures the dynamics of the swinging phase of the gait cycle. The linear leg model is defined by its moment of inertia $\mathbf{I_h}$, damping coefficient $\mathbf{b_h}$ and stiffness coefficient $\mathbf{k_h}$, while the corresponding parameters $\{\mathbf{I_e},\mathbf{b_e},\mathbf{k_e}\}$ are used to define the exoskeleton model as shown in Figure 2. The coupling between the leg and the exoskeleton is not rigid by virtue of the muscles, tissues, and structural and joint compliance of the exoskeleton. To model this, we use a linear spring-damper with damping coefficient $\mathbf{b_c}$ and stiffness coefficient $\mathbf{k_c}$ as shown in Figure 2.

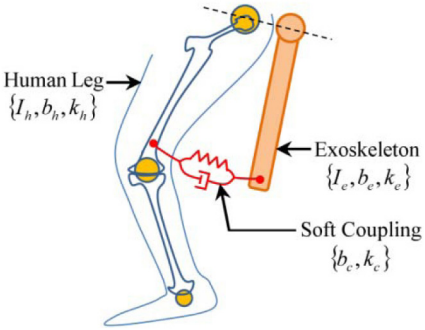

Figure 2. Coupled human-exoskeleton system.

The linear equations of motion of the coupled human-exoskeleton system are given by

$$I_h\ddot{\theta}_h(t) + b_h\dot{\theta}_h(t) + k_h\theta_h(t) = \tau_h(t) - \tau_c(t) \tag{1}$$

$$I_e\ddot{\theta}_e(t) + b_e\dot{\theta}_e(t) + k_e\theta_e(t) = \tau_e(t) + \tau_c(t) \tag{2}$$

$$b_c\left(\dot{\theta}_h(t) - \dot{\theta}_e(t)\right) + k_c\left(\theta_h(t) - \theta_e(t)\right) = \tau_c(t) \tag{3}$$

Here, τ_h and τ_e are the hip joint torques of the human and exoskeleton respectively, and θ_h and θ_e are the human and exoskeleton hip joint angles respectively. The coupling torque at the hip joint corresponding to the interaction between the human leg and the exoskeleton is denoted by τ_c.

The impedance [13] $Z_h(s)$ and admittance [13] $Y_h(s)$ transfer functions of an isolated human leg are given by

$$Z_h(s) = \frac{I_h s^2 + b_h s + k_h}{s}, \tag{4}$$

$$Y_h(s) = \frac{s}{I_h s^2 + b_h s + k_h}. \tag{5}$$

In this work, we use the integral of the admittance, called *integral admittance*, to characterize the joint dynamics. The integral admittance transfer function $X_h(s)$ is given by

$$X_h(s) = \frac{1}{I_h s^2 + b_h s + k_h}. \tag{6}$$

The admittance transfer function $Y_e(s)$ of the isolated exoskeleton and the impedance transfer function $Z_c(s)$ of the isolated coupling element are given by

$$Y_e(s) = \frac{s}{I_e s^2 + b_e s + k_e} \tag{7}$$

$$Z_c(s) = \frac{b_c s + k_c}{s} \tag{8}$$

Using (5), (7), (8), the equations of motion in (1), (2), (3) with an exoskeleton control transfer function $U_e(s)$ can be represented as a block diagram shown in Figure 3(*a*).

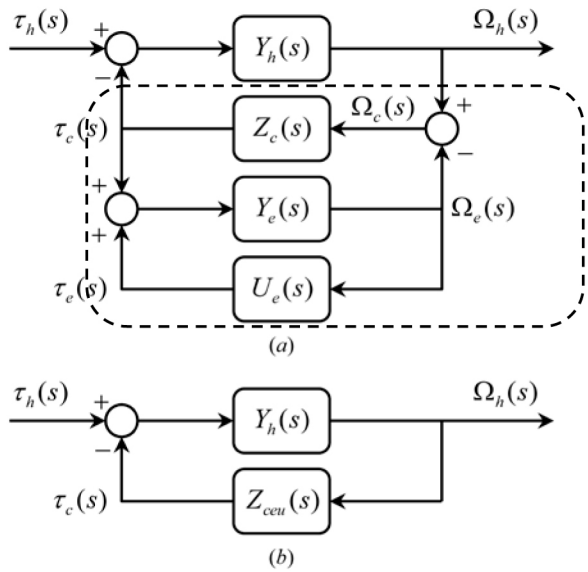

Figure 3. Block diagram of the closed-loop coupled human-exoskeleton system.

The outlined region in Figure 3(a) can be reduced to a single transfer function $Z_{ceu}(s)$ given by

$$Z_{ceu}(s) = \frac{Z_c(s)}{1 + Z_c(s)\,Y_{eu}(s)} \tag{9}$$

as shown in Figure 3(b). The transfer function $Y_{eu}(s)$ in (9) is given by

$$Y_{eu}(s) = \frac{Y_e(s)}{1 - Y_e(s)\,U_e(s)}. \tag{10}$$

The impedance transfer function $Y_{heu}(s)$ of the closed-loop system in Figure 3(b) is given by

$$Y_{heu}(s) = \frac{Y_h(s)}{1 + Y_h(s)\,Z_{ceu}(s)}, \tag{11}$$

and its integral admittance transfer function $X_{heu}(s)$ is given by

$$X_{heu}(s) = \frac{X_h(s)}{1 + Y_h(s)\,Z_{ceu}(s)} \tag{12}$$

The loop transfer function $L_{heu}(s)$ needed to evaluate the stability of the feedback system in Figure 3(b) is given by

$$L_{heu}(s) = Y_h(s)\, Z_{ceu}(s) \tag{13}$$

and its gain margin is given by

$$GM(L_{heu}) = \frac{1}{|L_{heu}(j\,\omega_c)|}, \tag{14}$$

where ω_c is the phase-crossover frequency when the phase of $L_{heu}(s)$ is 180°, i.e., $\angle L_{heu}(j\omega_c) = 180°$. In order for the closed-loop system to be stable, the gain margin in (14) must satisfy the following condition:

$$GM(L_{heu}) > 1. \tag{15}$$

Additionally, the closed-loop system must satisfy the following phase condition

$$\angle X_{heu}(j\omega) \in [-180°, 0°], \qquad \forall\omega \tag{16}$$

in order to be passive [12]. Passivity ensures that the human leg with the exoskeleton does not become unstable when it contacts passive environments like the ground.

INTEGRAL ADMITTANCE SHAPING

The integral admittance $X_{heu}(s)$ in (12) defines the linear dynamics of the closed-loop coupled human-exoskeleton system. The isolated human's integral admittance $X_h(s)$ in (6) is modified to the $X_{heu}(s)$ in (12) by virtue of the coupling, exoskeleton and its controller $U_e(s)$. The controller can be chosen such that the desired dynamics can be achieved. Figure 4 shows the magnitude profile of the frequency response of $X_h(s)$ and a hypothetical $X_{heu}(s)$ for a coupled human-exoskeleton system whose parameters are shown in Table 1.

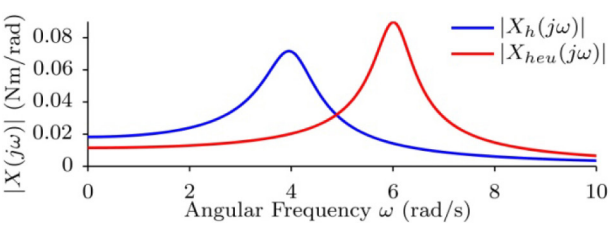

Figure 4. Integral admittance frequency response magnitude profiles.

Every dynamic response corresponds to a particular shape of the frequency response magnitude profile $|X_{heu}(j_\omega)|$, one example of which is shown in Figure 4. But what should the desired dynamics be? In the following section, we take an intuitive approach to answering this question.

Table 1. Human leg, exoskeleton and coupling parameters for a human subject 165 cm tall weighing 65 kg.

Human leg length l_h	0.875 m
Human leg moment of inertia I_h	3.381 kg m^2
Human hip joint damping coefficient b_h	3.5 Nms/rad
Human hip joint stiffness coefficient k_h	54.677 Nm/rad
Exoskeleton moment of inertia I_e	0.01178 kgm^2
Exoskeleton joint damping coefficient b_e	0.34512 Nms/rad
Exoskeleton joint stiffness coefficient k_e	0.33895 Nm/rad
Coupling joint damping coefficient b_c	9.474 Nms/rad
Coupling joint stiffness coefficient k_c	1905.043 Nm/rad

Assistance and Resistance Defined

It can be easily understood that a lighter leg will be easier to move than a heavier one. So, the human leg can be moved with less effort if its moment of inertia I_h can be reduced. Similarly, reducing the damping coefficient b_h and stiffness coefficient k_h at the joint will also tend to make it easier for the human to move his/her leg. Thus, in order to provide assistance, the exoskeleton controller $U_e(s)$ must be chosen such that the moment of inertia, damping coefficient and stiffness coefficient of the dominant second-order dynamics of the coupled human-exoskeleton system must be lowered below that of the unassisted human leg. This translates to reducing the impedance and increasing the admittance of the coupled human-exoskeleton system with respect to that of the unassisted human leg.

Thus, in this work, we define *assistance* as increasing the admittance of the coupled human-exoskeleton system with respect to that of the unassisted human leg. Similarly, *resistance* is defined as decreasing the admittance of the coupled system. More precisely, we consider a coupled system to be *assisted*, if its frequency response magnitude is greater than that of the unassisted human leg for all frequencies. Figure 5 highlights the range of frequencies where assistance and resistance are experienced for the hypothetical coupled system presented in Figure 5.

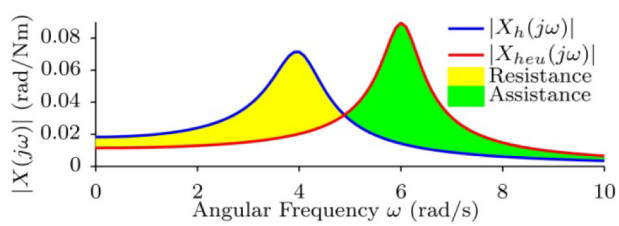

Figure 5. Assistance and resistance defined.

Now that we have defined assistance based on the frequency response magnitude profile of the integral admittance of the coupled system, let's study how the shape of the frequency response magnitude profile changes while reducing the system parameters. Figure 6 graphically shows how the integral admittance frequency response magnitude profile changes for three different cases: (i) when I_h is halved, (ii) when b_h is halved and (iii) when k_h is halved. In each case, the other parameters were kept fixed at their nominal values.

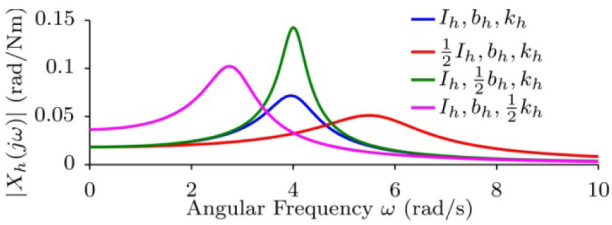

Figure 6. Effect of changing the system parameters on the integral admittance frequency response magnitude profile.

Reducing the Moment of Inertia

We see in Figure 6 that reducing the moment of inertia results in a frequency response magnitude profile, which is higher than that of the unassisted human leg for higher frequencies and lower for lower frequencies. This implies that at higher input frequencies, the integral admittance of the coupled system will be higher than that of the unassisted human leg, and hence assistive as desired. However, at lower input frequencies, the integral admittance of the coupled system will be lower, and hence resistive, which is not desired. Therefore, reducing the moment of inertia of the coupled system below that of the unassisted human leg will be assistive at higher frequencies and resistive at lower frequencies.

Reducing the Damping Coefficient

Figure 6 shows that reducing the damping coefficient results in a magnitude profile that is higher than that of the unassisted human leg around its peak resonant frequency. However, no discernible difference is observed at lower and higher frequencies. Reducing the damping coefficient might be a good starting point towards developing assistive controllers as it produces no discernible resistance at any frequency and provides assistance around the peak resonant frequency.

Reducing the Stiffness Coefficient

Reducing the stiffness coefficient results in an integral admittance magnitude profile that is higher than that of the unassisted human leg at lower frequencies as shown in Figure 6. This implies that the human feels assisted at lower frequencies, which is desired. However, at higher frequencies, the human experiences resistance as the integral admittance magnitude is lower than that of the unassisted human leg. Therefore, reducing the stiffness coefficient of the coupled system below that of the unassisted human leg will be assistive at lower frequencies and resistive at higher frequencies, which is opposite to the effect of reducing the moment of inertia.

Reducing the Impedance

Figure 6 and the above sections described the effect of independently reducing the moment of inertia, damping coefficient and stiffness coefficient while keeping the other parameters constant. None of the cases produced assistance at all frequencies. Reducing moment of inertia, reducing stiffness coefficient and reducing damping coefficient increased the integral admittance magnitude at almost disjoint set of frequencies as can be seen in Figure 6. So, how about reducing all the parameters together? Can this potentially achieve an integral admittance magnitude profile that is higher than that of the unassisted human leg at all frequencies like the hypothetical one shown in Figure 7?

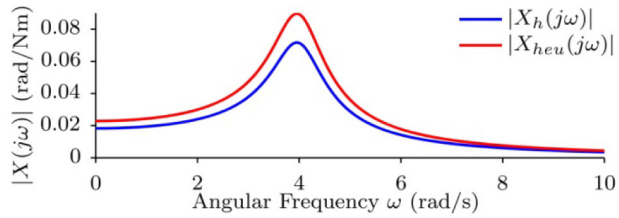

Figure 7. A hypothetical coupled human-exoskeleton system that is assisted at all frequencies.

The answer to the above question is in the affirmative. Indeed, all the system parameters can be potentially reduced in order to achieve assistance at all frequencies. The exoskeleton controller $U_e(s)$ can be chosen such that the moment of inertia, damping coefficient and stiffness coefficient of the dominant second-order dynamics of the coupled human-exoskeleton system is potentially lowered below that of the unassisted human leg.

Before proceeding to present an exoskeleton control law that achieve assistance at all frequencies, we present the following quantitative metrics for assistance and resistance, namely *assistance ratio A* and *resistance ratio R*, that will be used to find the exoskeleton control parameters. These metrics are defined based on the frequency regions highlighted in Figure 5 and exploits the fact that at any particular frequency ω, the coupled system experiences either assistance or resistance but not both.

Assistance ratio A

$$= \begin{cases} \dfrac{1}{\omega_f}\displaystyle\int_0^{\omega_f} \dfrac{|X_{heu}(j\omega)| - |X_h(j\omega)|}{|X_h(j\omega)|}\, d\omega & if\ |X_{heu}| \geq |X_h| \\ 0 & if\ |X_{heu}| < |X_h| \end{cases} \tag{17}$$

Resistance ratio R

$$= \begin{cases} 0 & if\ |X_{heu}| \geq |X_h| \\ \dfrac{1}{\omega_f}\displaystyle\int_0^{\omega_f} \dfrac{|X_h(j\omega)| - |X_{heu}(j\omega)|}{|X_h(j\omega)|}\, d\omega & if\ |X_{heu}| < |X_h| \end{cases} \tag{18}$$

Exoskeleton Control

How can the integral admittance of the coupled human-exoskeleton system be increased with respect to the unassisted human leg? It can be achieved using a powered exoskeleton with an appropriate control law. In this work, we use an exoskeleton controller that positive

feedback on hip joint angle and joint angular velocity. It also has positive feedback on hip joint angular acceleration filtered by a low pass filter as shown in

$$U_e(s) = \frac{K_\alpha H_{lo}(s)s^2 + K_\omega s + K_\theta}{s}, \tag{19}$$

where $H_{lo}(s)$ is the second-order low-pass Butterworth filter with a cut-off frequency ω_{lo} given by

$$H_{lo}(s) = \frac{\omega_{lo}^2}{s^2 + \sqrt{2}\omega_{lo}s + \omega_{lo}^2}. \tag{20}$$

The positive angle feedback using gain K_θ and positive angular velocity feedback using gain K_ω correspond to reducing the stiffness and damping coefficients respectively. The positive angular acceleration feedback using gain K_α corresponds to reducing the moment of inertia. However, a low-pass filter is used for acceleration feedback in order to provide more stability while reducing the moment of inertia. More detailed explanation for the use of a low-pass filter is beyond the scope of this paper.

The exoskeleton controller $U_e(s)$ has four parameters, namely K_θ, K_ω, K_α and ω_{lo}, which can be tuned such that the desired assistance ratio A_d is achieved. Since the exoskeleton controller uses positive feedback, stability of the coupled system is one of the main concerns. So, while attempting to achieve the desired assistance ratio, we should also ensure that coupled stability and passivity are guaranteed, i.e., the conditions in (15) and (16) are satisfied.

The problem of finding the optimal control parameters that achieve the desired assistance ratio A_d along with negligible resistance ratio, if any, and satisfy the coupled stability and passivity constraints can be formulated in to a constrained optimization problem as follows:

$$\begin{aligned} &\underset{\{K_\theta, K_\omega, K_\alpha, \omega_{lo}\}}{minimize} && |A - A_d|^2 + wR \\ &subject\ to && GM(L_{heu}) > 1 \\ & && \angle X_{heu}(j\omega) \in [-180°, 0°], \ \forall\omega \\ & && \left|\frac{\zeta_{heu} - \zeta_h}{\zeta_h}\right| < \epsilon \end{aligned} \tag{21}$$

The above optimization problem has a constraint on the damping ratio ζ_{heu} of the coupled system in addition to the stability and passivity constraints. The constraint demands the damping ratio ζ_{heu} to remain in the ε-neighborhood of that of the unassisted human leg ζ_h so as to make the human feel more comfortable. The objective function uses a large weight w on the resistance ratio R in order to keep it as low as possible while achieving the desired assistance ratio A_d.

Optimization Results

Figure 8 presents the frequency response magnitude profiles of the integral admittance of the coupled human-exoskeleton system for desired assistance ratios of 0.05 and 0.10, and the zero desired assistance ratio profile corresponds to that of the unassisted human leg. The exoskeleton control parameters were optimized for each of these cases using the optimization in (21).

We see in Figure 8 that the profiles corresponding to $A_d = 0.05$, 0.10 show assistance at frequencies around the resonant peak frequency with a shift in the resonant peak frequency. Meanwhile, there is no discernible resistance at any frequency. Unlike with only reducing the damping coefficient, here, the resonant peak frequency increases with assistance, which corresponds to an increase in the natural frequency of the leg. This can potentially help the human take faster steps. For desired assistance ratio $A_d > 0.14$, the coupled system was stable but not passive, and hence larger desired assistance ratios were not attempted.

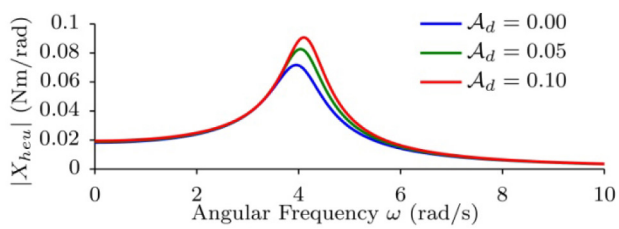

Figure 8. Optimized integral admittance shapes for different desired assistance ratios.

The Nyquist plots in Figure 9 demonstrate that the coupled systems corresponding to $A_d = 0.05$, 0.10 are both stable as there are no encirclements of $-1+j0$.

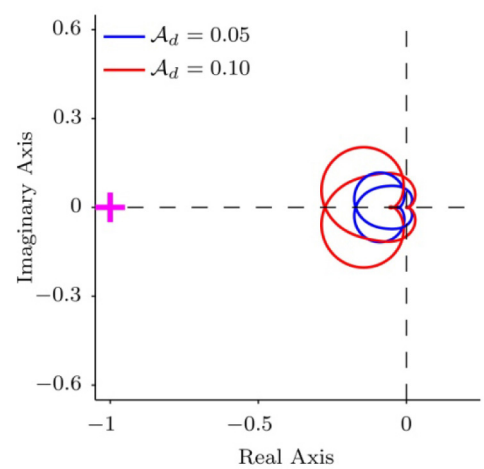

Figure 9. Nyquist plots for the coupled human-exoskeleton systems optimized for different desired assistance ratios indicating that they are stable.

EXPERIMENTAL RESULTS

We tested the integral admittance shaping control framework on the Honda Stride Management Assist (SMA) device shown in Figure 1. The SMA device has hall-effect sensors on the hip actuators, which provide exoskeleton angle θ_e measurements. However, there are no

direct measurements for the angular velocity $\dot{\boldsymbol{\theta}}_{\boldsymbol{e}}$ and angular acceleration $\ddot{\boldsymbol{\theta}}_{\boldsymbol{e}}$, and hence we use a Kalman filter for estimating these values as shown in the control block diagram in Figure 10. The exoskeleton control gains are obtained from the integral admittance shaping optimization in (21) for a desired assistance ratio of 0.05. For larger assistance ratios, the SMA motor torques saturated, and hence controllers corresponding to the larger assistance ratios were not tested.

The integral admittance shaping controller was evaluated on a human subject 165 cm tall weighing 65 kg. The human subject was asked to perform two activities, namely straight line walking and stair climbing, with and without the assist controller. Figures 11 and 12 show the phase plots of the hip joint motion during walking and stair climbing respectively. Note that in both cases, unassisted and assisted, the subject is wearing the device. In the unassisted case, the device has the assistive controller is turned off, while in the assisted case, it is turned on. It can be seen that for both activities, namely walking and stair climbing, the hip joint motion was amplified when our assist controller was used. Thus, these results experimentally validate the motion amplification characteristics of our integral admittance shaping controllers.

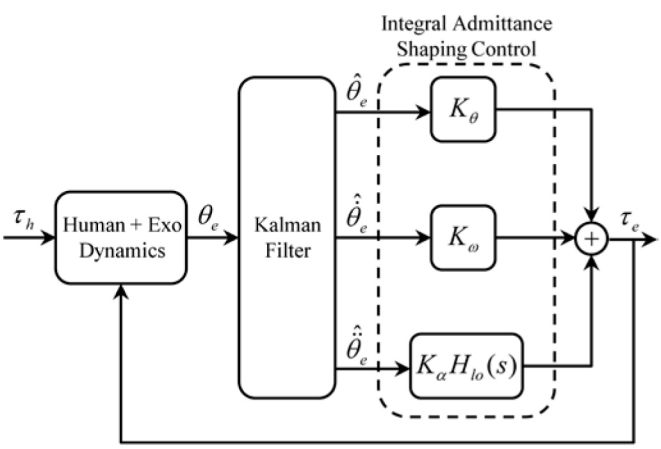

Figure 10. Exoskeleton control framework on Honda Stride Management Assist (SMA) device.

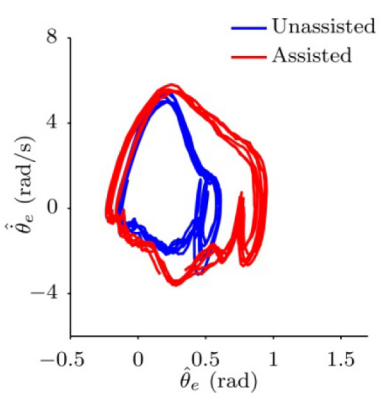

Figure 11. Phase plot of the hip joint motion during walking with and without the exoskeleton assist controller.

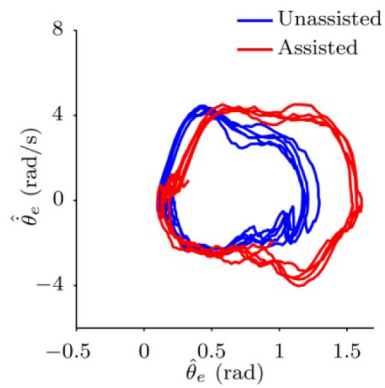

Figure 12. Phase plot of the hip joint motion during stair climbing with and without the exoskeleton assist controller.

CONCLUSIONS

In this paper, we have shown that a control approach based on the concept of active impedance can result in a general framework for assistive exoskeleton controllers. Although this framework, which we call *Integral Admittance Shaping*, is conceptually uncomplicated, it is surprisingly robust and is able to provide assist to a broad range of human activities of daily living (ADL). Importantly, the controller is *neutral* and it functions only in amplifying the "input" human motion. A neutral controller does not attempt to impose its own "control law" on the human; it behaves under the assumption that the human is the best judge in deciding what motion to adopt in a given situation. Otherwise, there would be the possibility of a conflict between the two feedback control laws, one that is inherent to the human, and the other that is adopted by the exoskeleton.

We have presented successful experimental results using human subjects engaged in two different activities, straight line walking and stair climbing. In both cases, the beneficial effect of assist is clear from the viewpoint of motion amplification.

Future work includes both theoretical exploration of the controller and human subject tests. A number of parameters define the assist controller reported in this paper. The parameter space is large and we have so far been able to explore only a limited zone of this space. A thorough exploration of the available parameter space will be worthwhile. Specifically, the effect of different parameters on human, i.e., what type of assist does a human subject feel for different combinations of the parameters will be instructive.

REFERENCES

1. Ageing and life-course, World Health Organization, www.who.int/ageing/en/.
2. Hartholt, K.A., Stevens J.A., Polinder S., van der Cammen, T.J. et al., "Increase in fall-related hospitalizations in the United States, 2001-2008", J. Trauma, 71(1):255-8, 2011, doi:10.1097/TA.0b013e31821c36e7.
3. Falls, Fact sheet N°344, World Health Organization, http://www.who.int/mediacentre/factsheets/fs344/en/.
4. Dollar, A.M., and Herr, H., "Lower extremity exoskeletons and active orthoses: Challenges and state-of-the-art," IEEE Trans. Robotics, 24(1):144-158, 2008, doi:10.1109/TRO.2008.915453.
5. Ferris, D.P., "The exoskeletons are here," Journal of Neuroengineering and Rehabilitation, 6(17):1-3, 2009.

6. Ferris, D.P., Viant, T.L., and Campbell, R.J., "Artificial neural oscillators as controllers for locomotion simulations and robotic exoskeletons", 4th World Congress of Biomechanics, 2002, doi:10.1186/1743-0003-6-17.
7. Ronsse, R., Koopman, B., Vitiello, N., Lenzi, T., et al., "Oscillator-based walking assistance: a model-free approach", IEEE Int. Conf. Rehabil. Robot. (ICORR), 668-674, 2011, doi:10.1109/ICORR.2011.5975352.
8. Van der Pol, B., "On relaxation-oscillations", The London, Edinburgh and Dublin Pil. Mag. & J. of Sci., 2(7):978-992, 1926.
9. Duffing G., "Erzwungene Schwingungen bei Veränderlicher Eigenfrequenz.", F. Vieweg u. Sohn, Braunschweig, 1918.
10. Hooper, S.L., "Central Pattern Generators", eLS, John Wiley & Sons, 2001, doi:10.1038/npg.els.0000032
11. Honda Stride Management Assist (SMA) device, http://corporate.honda.com/innovation/walk-assist/.
12. Colgate, J.E., and Hogan, N., "An analysis of contact instability in terms of passive physical equivalents," in Proc. IEEE Int. Conf. Robot. Autom. (ICRA), 404-409, 1989.
13. Hogan, N. and Buerger, S.O., "Impedance and Interaction Control", Robotics and Automation Handbook. CRC Press, 2005.

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DEFINITIONS/ABBREVIATIONS

ADL - Activities of Daily Living

DOF - Degree of Freedom

SMA - Stride Management Assist

WHO - World Health Organization