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Ankle controller design for robotic ankle-foot prostheses to reduce balance-related effort during walking using a dynamic walking approach

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Abstract

The goal of my research is to develop ankle-foot prosthesis controllers that reduce balance-related effort during walking. Although great progress has been made in ankle foot prostheses, individuals with below knee amputation still report difficulty with balance. While the effective balancing method of foot placement is unavailable in ankle-foot prosthesis, the balance restoring resource of ankle actuation holds potential for amputee walking. I explored the possibilities for prosthetic foot designs to improve balance through simulation studies, hardware development, and human subject experiments. I demonstrate that ankle actuation control can be very important in balance maintenance, and present two new approaches to reduce balance-related effort for people with lower limb amputations.

Through a simulation of three-dimensional limit cycle walking of amputee gait, I demonstrate that ankle actuation can be equally effective as foot placement, especially in once-per-step modulation of ankle push-off work. I implemented the ankle push-off work controller in an ankle-foot prosthesis emulator and tested the controller on human subjects. I found that with this push-off work controller, both able-bodied subjects with simulated amputation and individuals with below knee amputation reduced balance-related effort. One possible explanation of amputee's reduced metabolic rate could be their reduced intact limb control effort during stance phase. In addition, more training seemed to help amputee participants realize the benefits of the controller.

Simulation results also suggest that inversion/eversion control could improve balance. To test control ideas, I developed a two degree-of-freedom ankle-foot prosthesis with plantarflexion and ankle inversion/eversion. Using this device, I

investigated the balancing effect of passive ankle inversion/eversion stiffness and active once-per-step modulation of inversion/eversion torque. The inversion stiffness strongly affected amputee's balance-related effort. Active inversion controller lowered metabolic rate, a balance-related effort indicator. While these step-to-step variations in ankle/inversion torque reduced balance-related efforts, these effects were not as effective as those of the ankle push-off work control.

The results from these simulation studies and human experiments suggest that step-to-step alteration in ankle actuation can reduce balance-related effort. This finding will help inform future design of prosthetic devices, which could reduce balance-related effort, increase balance confidence, and improve overall quality of life.

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Chapter 1

Introduction

1.1 Motivation

Walking with amputated legs is associated with several deficits, including increased balance-related effort to walk, more frequent falling incidents, decreased balance confidence, avoidance of social activity, reduced mobility, and reduced quality of life. I try to solve some of these problems by providing an active ankle-foot prosthesis using a dynamic walking approach.

One out of every two hundred Americans is estimated to have limb amputation (Adams et al., 1999). In 2005, ten million people were estimated to have below knee amputation (Ziegler-Graham et al., 2008). Losing a lower limb causes several deficits including increased falling incidents, fear of falling, and balance-related effort. Approximately 50% of amputees experience falling incidents (Miller et al., 2001b) that could result in severe bodily injury (Gonzalez and Mathews, 1980; Miller et al., 2001b) and cost more than a billion medical dollars, based on an estimation using previous report of cost per fall in dollars (Shumway-Cook et al., 2009). This deficit might be related to reduced balance ability (Gates et al., 2013b; Segal and Klute, 2014; Paysant et al., 2006; Muir et al., 2010b,a), as the individuals who took particular effort at each step reported an increased fear of falling (Miller et al., 2001b). Reduced balance may also partially contribute to an increase in metabolic energy consumption (Waters and Mulroy, 1999; Paysant et al., 2006), similar to when able-bodied subjects increase their consumption to maintain balance when their sensory information has been disturbed (Voloshina et al., 2013; O'Connor et al., 2012). Those balance-related deficits are associated with reduced balance confidence (Miller et al., 2001a, 2002), which seems to result in avoiding social activity, a reduced mobility, and reduced quality of life (Miller et al., 2001a). These deficits may lead them to want a device to assist with balance during walking (Legro et al., 1999; Hagberg and Brånemark, 2001).

Several intervention methods have been developed to help individuals with below knee amputation. A method used in nursing homes has been modifying the environment (Tideiksaar et al., 1993). Wards have been educated to watch patients more carefully to avoid falling accidents (Gooday and Hunter, 2004). Although these methods have been effective at reducing the falling rate or consequences of falling, they are ineffective when patients want to walk independently, especially outside the home. To address this issue, several training methods have been developed, such as an increased usage of the hip muscles (Heidi Nadollek and Isles, 2002; Esquenazi and DiGiacomo, 2001), gait training (Crenshaw et al., 2013a), task-oriented balance training (Kaufman et al., 2014), prosthesis usage training (Kulkarni et al., 1996) and teaching a method to fit a socket to enhance balance by improving sensory feedback from a stump (Vittas et al., 1986). Some of these methods yielded positive results in balance confidence and falling occurrences. However, these methods may not fully meet the amputee's need for a device that allows them to walk easier by assisting balance (Legro et al., 1999; Hagberg and Brånemark, 2001).

Researchers have started to study a balance assisting feature in ankle-foot prosthesis by incorporating compliance (Lindhe, 2014) or semi-active actuation (Panzenbeck and Klute, 2012). Those devices seem to be promising (Panzenbeck and Klute, 2012), but the efficacy in balance-related effort is inconclusive (Childers et al., 2015; Segal and Klute, 2014). On the other side, researchers explored an active ankle-foot prosthesis to reduce overall effort during walking (Herr and Grabowski, 2012). Although this device showed a reduction in overall energy consumption, the effect of such active devices on balance is not clear (Gates et al., 2013a). Perhaps, by designing active controllers that specifically consider balance-related effort, such effort could be reduced during walking, and walking might become even easier.

Balance-related effort could be reduced by providing control action at each step while maintaining average behavior. Step-to-step adjustments have been shown to maintain balance during walking for able-bodied subjects (Collins and Kuo, 2013; Voloshina et al., 2013; O'Connor et al., 2012) and amputees (Gates et al., 2013b) especially when balance has been challenged (Voloshina et al., 2013; O'Connor et al., 2012; Gates et al., 2013b). Individuals with below knee amputation also seem to exert more discrete (Gates et al., 2013b) and continuous (Hof et al., 2010) step-to-step correction than their able-bodied counter parts including foot placement and intact limb control, perhaps to compensate for their loss of ankle actuation. In robotics, this type of step-to-step control method successfully stabilized many walking robots (Hobbelen and Wisse, 2008; Bhounsule et al., 2012), including a robot that set the record for longest walking distance (Bhounsule et al., 2012). This suggests that an ankle-foot prosthesis providing step-to-step correction might help amputees walk with less active control effort.

For a robotic prosthesis, ankle plantarflexion and ankle inversion/eversion can be used to restore balance during walking. Ankle actuations seem to be related to balance, as shown by the resultant increased balance-related effort when the actuation is lost (Paysant et al., 2006). Controlling ankle push-off work via ankle plantarflexion control seems to be an effective method to stabilize two-dimensional walking (Hobbelen and Wisse, 2008; Bhounsule et al., 2012), although the applicability to three-dimensional walking has not been investigated. Ankle inversion/eversion actuation seems to help restore balance after foot placement (Hof et al., 2010, 2007), especially on the intact limb side for individuals with leg amputations (Hof et al., 2007). If those two control methods are relatively important compared to other control strategies in three-dimensional walking, then we can give a benefit by providing an active ankle foot prosthesis. Simulation studies may be useful to provide controlled comparison results of the effect of different actuation strategies on stability.

The aim of this work is to develop controllers for a robotic ankle-foot prosthesis to reduce balance-related effort. I hypothesized that step-to-step adjustment in plantarflexion and inversion/eversion would reduce such an effort in an ankle-foot prosthesis. For this end, I developed a three-dimensional numerical model of amputee gait for the purpose of generating and exploring novel control ideas that enhance balance during walking. Based on our simulation results, I implemented different controllers on ankle-foot prosthesis emulators and evaluated their efficacy by performing human-subject experiments. In the course of this development, I also developed two degrees of freedom ankle-foot prosthesis to explore full possibility of ankle actuation. In the next section, I review related work.

1.2 Background

1.2.1 Limit cycle walking

Researchers often utilize the limit cycle walking principle to find natural and efficient walking motions and related control methods. At each moment in time, the limit cycle walking is not stable in isolation. However, when examined over a course of multiple steps, the walking becomes stable. Compared to other walking methods, such as zero moment point control, the limit cycle walking principle permits a more realistic and efficient walking motion because it introduces fewer constraints by creating stability using several steps, rather than a single step.

Stability of limit cycle walking

By examining step-by-step, the stability of limit cycle walking can be deduced. If the limit cycle is stable, than the nominal trajectory will be neared by the adjacent trajectories as several steps are taken (cyclic stability). A point within a step can be mapped onto another point in a subsequent step via stride function (McGeer, 1990). If the mapped point is identical to the initial point, then a periodic motion exists. This point is called a fixed point. By linearizing the stride function about the fixed point, a linearized matrix can be obtained. If the matrix's eigenvalues, also referred to as Floquet Multipliers, are within the unit circle, then the walking motion is stable (Hurmuzlu and Moskowitz, 1986).

Dynamic walking

An analysis of dynamic walking can utilize the limit cycle walking principle. Dynamic walking is a type of passive dynamic walking with carefully added actuations to stabilize walking motion. A passive dynamic walker is cyclically, rather than locally stable, meaning that while the model can complete several strides without compromising stability, they are not able to stand still. This model has features of energy efficient and realistic walking motion, which has been used to explain several characteristics of human walking motion (Kuo et al., 2005; van der Krogt et al., 2010). Another feature of this model is the small number of parameters needed to generate walking motion. This aspect allows for thorough investigation of the parameter space and assists in the comprehension of walking motion. The passive dynamic walker can recover from larger disturbance through the addition of a controller or actuator, which creates a dynamic walking model. If the supplemental actuations mostly allow dynamic motion, then many similar traits will remain such as cyclical stability. Dynamic walking motion has been used to explain human walking attributes such as energetics (Kuo, 2001), efficient stabilization method (Kuo, 1999; Bauby and Kuo, 2000), and balance-related energy cost (Donelan et al., 2004; IJmker et al., 2013; Dean et al., 2007). This model also can be used to formulate controllers (Hobbelen and Wisse, 2008; Bhounsule et al., 2012). Likewise, similar models can be used to understand the actuation role in stability and to design a controller to reduce balance-related effort.

1.2.2 Stabilization methods

Unlike two-dimensional passive dynamic walking models, where self-stabilization may be effected through the cancellation of collision by push-off work, a three-dimensional walking model usually needs a control to prevent collapses (Kuo, 1999). Similarly, active control seems to be necessary to maintain individuals balance especially in the medio-lateral direction (Donelan et al., 2004; Bauby and Kuo, 2000).

The medio-lateral motion during walking can be energy efficiently stabilized using a foot placement strategy via hip joint movement (Kuo, 1999). Similar technique seems to be used in human walking for able-bodied participants (Collins and Kuo, 2013; Voloshina et al., 2013; Donelan et al., 2004; O'Connor et al., 2012) or individuals with below-knee amputation (IJmker et al., 2013). This correction method is, however, difficult to utilize in a robotic ankle foot prosthesis.

One feasible strategy is a center of pressure control. This strategy has been widely used to stabilize walking motion. Center of pressure control occurs after the foot placement control decides the possible center of pressure region. This control can further correct the center of pressure location to stabilize walking motion, especially in the medio-lateral direction. One successful approach in humanoids is the use of ankle inversion/eversion actuation (Kim et al., 2007). Humans also seem to adopt this control technique to some degree (Hof et al., 2010), especially people with leg amputations (Hof et al., 2007). While this control strategy is a possible solution for lateral stabilization within an active ankle-foot prosthesis, because of finite foot width and an under-actuation phase, the center of pressure control strategy is difficult to implement in the prosthesis.

Ankle push-off work control could be an effective method if it is able to stabilize walking motion in the medio-lateral direction as well as in the fore-aft direction. In the fore-aft direction, push-off work can stabilize walking by removing/adding energy from/to a model when the model is disturbed and has increased/decreased energy. For instance, when a model has a short step length due to a disturbance, the model dissipates less energy by collision. The model then needs less energy in the next step, which can be fulfilled by decreasing the amount of push-off work. However, it remains unclear if medio-lateral direction can be similarly stabilized by push off work modulation. If ankle push-off has control authority in the medio-lateral direction, then we can provide a benefit by using a robotic ankle foot prosthesis. An investigation using a limit cycle walking model might elucidate this issue.

In a simulation study, controlling hip and ankle actuations at each step can stabilize a three-dimensional limit cycle model of walking. More specifically, a limit cycle walking model can be stabilized by changing actuation parameters including step width, ankle inversion/eversion resistance, and push-off work, once-per-step. A linear feedback control scheme can be applied to generate the input parameter of actuation. More specifically, a controller samples states at a fixed point, and uses this information to compute a nominal state error, which, when multiplied by a gain matrix, produces the control input used in the next step

A gain matrix can be calculated in several ways. One method uses a linear quadratic regulator, which optimizes the cost function by considering state errors and control inputs. The cost function can be written as a quadratic function with appropriate weights on states and control inputs. A linear differential equation for the function can be made by linearizing the stride function at a fixed point. Then, an optimal solution can be determined by solving the Ricatti equation. This method proves its usefulness especially when the desired pole location is unknown and balance between state errors and control inputs is necessary.

The gain matrix can be optimized further for cases outside the linear region by using a different optimization method such as a covariance matrix adaptation evolutionary strategy (CMA-ES) (Hansen and Kern, 2004). This algorithm renews the mean and covariance matrix of distribution to increase the likelihood of success of search step and previous candidate solution. Because the algorithm can find near global optimum solutions with small manipulations, we can analyze the stabilizing effects of each actuation more thoroughly.

1.2.3 Balance-related measures

A stable walking motion would not involve falling. Compared to a less stabilizing controller, a more stabilizing controller would allow a model to walk without falling under larger disturbances. An ideal balance-related measure may capture this likelihood of falling under a certain disturbance. Several measures exist for a simulation study to measure stability including maximum tolerable disturbance in random ground height before the model falls down. In an experimental study, it would require lots of attention to make a person fall down and use it as a measure. Still, several other alternative methods exist for measuring the balance-assisting performance of a controller.

In a simulation study, we could use several stability measures to estimate the balance-restoring capacity of a controller but many of them are not directly related to the likelihood of falling (Bruijn et al., 2013). The maximum Floquent multiplier is often used to analyze stability. It measures a disturbance rejection rate. This measure is applicable for a linear region, but usually a small disturbance easily moves the state to a nonlinear region. In addition, the direction of a disturbance might be different from the most unstable direction within a linear region. These characteristics make this multiplier a poor indicator of the likelihood of falling. Basin of attraction measures stability by considering a nonlinear region. Using such a measure, the maximum allowable disturbance can be measured, but the direction could be hard to estimate and computationally expensive. The gait sensitivity norm can capture the rate of disturbance rejection and determine the allowable disturbance with a very carefully selected gait indicator to measure stability meaningfully (Hobbelen and Wisse, 2007). The maximum allowable disturbance considers a nonlinear region, has a direction, does not require careful selection of an indicator, and is directly related to the likelihood of falling (Song et al., 2013). This measure still needs to decide how many steps a model needs to walk to robustly measure the stability before increasing the level of disturbance.

In addition to the stability measures, the energy requirement for each controller can provide additional stabilization performance in terms of efficiency when a model walks under disturbed conditions. A good candidate for a stabilization controller may provide reasonable stabilizing performance with low energy expenditure by the controller.

In a human-subject experiment, measuring balance-related effort might be a practical method for examining the balance-assisting performance of a specific controller. The balance-related effort indicates the active control effort of a subject during walking to maintain balance, including step-to-step control effort or nominal control effort if a controller is being used to maintain balance.

One of the widely used measures is step width variability, which captures active foot placement control effort under disturbance (Voloshina et al., 2013; O'Connor et al., 2012). The variability and metabolic energy consumption are reduces by an assistive device (Dean et al., 2007; Donelan et al., 2004; IJmker et al., 2013), possibly indicating that subjects rely more on natural swing dynamics by using the device. This suggests that differences in step width variability may indicate the effect of each controller on the balance assisting performance.

Step width average also can indicate a difference in the balance-related effort of a subject. Step width affects the margin of stability with a cost of metabolic energy consumption. Humans tend to use energy to compensate for reduced balance by increasing step width when there is a balance-related deficit (Curtze et al., 2011) or when they are exposed to a challenging environment (Voloshina et al., 2013). To minimize the effort to maintain balance while walking, subjects might reduce their step width average if a device could help them to restore balance.

The center of pressure variability would indicate that the ankle inversion/eversion control effort after foot placement control has been finished to maintain a balance during walking (Hof et al., 2010). A higher center of pressure variability was observed on the intact limb side of individuals with leg amputation (Hof et al., 2007). This higher variability might suggest that with reduced balance capability, subjects might use the available balance strategy more actively, which in this case is ankle inversion/eversion. By providing a balance assisting device, subjects might reduce the inversion/eversion control effort.

Metabolic energy consumption may capture the altered muscle activity needed to maintain balance (Voloshina et al., 2013). Walking in the face of disturbance increases metabolic energy consumption (Paysant et al., 2006; O'Connor et al., 2012; Voloshina et al., 2013), whereas providing an external stabilizer reduces it (Donelan et al., 2004; IJmker et al., 2013). Similarly, stabilizing ankle-foot prosthesis controllers might reduce the energy consumption.

Active balance control may require cognitive load and can be revealed in dual tasks. For instance, concentrating more on gait has resulted in low gait variability but reduced the dual task score (Hollman et al., 2007). Different performances of stabilizing controllers may result in either differences in gait variability measures or differences in dual task scores.

Finally, user preference might show the effects of different controllers on balance assisting capability. Users preferred an actively controlled prosthetic knee (Stinus, 2000; Kaufman et al., 2007), which seems to reduce balance-related effort and falling rate (Highsmith et al., 2010). Similarly, users might prefer a controller that positively influences balance. A combination of these balance-related effort measures can show how effectively a controller reduces the balance-related effort of a subject.

1.2.4 Experimental design

To supplement the traditional group research methods, single-case studies were also used in some studies to investigate each individual's response considering high inter-subject variability among individuals with below knee amputation.

In group research, a hypothesis can be evaluated by examining group response using a between-group design or repeated measures deign. The between-group design is an experimental protocol that randomly tests different factors simultaneously using two or more groups. In repeated measured design experiments, each subject is exposed to all conditions. After conducting experiments, the hypothesis can be tested by comparing group means with statistical analysis with statistical assumption. Group research allows detection of weak effects and interactions between intervention and subjects. However, using group research to detect the efficacy of the controller may be less effective when high inter-subject variability exists because the average response can mask the efficacy of an intervention. For such a case, probing each individual's response may reveal the effect of given intervention.

Single-case research

Single-case experiments investigate the effectiveness of intervention for a single individual. This research method tries to find an intervention that significantly improve each individual's behavior. Therefore, this design might be considered as more clinically useful. Some typical designs are as follows. Usually, in the beginning, a baseline behavior is determined based on observation (Dermer and Hoch, 2012). After establishing the baseline, behavior is beneficially altered through the introduction of an intervention. Such a study, which consists of a baseline and intervention phase, is considered an A-B design. An A-B-A design involves removal of the intervention, which can evaluate repeatability of the intervention and also can test long-term effect of the intervention. However, this design can also prove unethical if the participant's condition worsens in the absence of the intervention. In an attempt to assuage the detriments and utilize the benefits of the A-B-A design, an A-B-A-B design was developed. This design involves the removal of an intervention temporarily, but is followed by the intervention's reintroduction if needed to further improve behavior.

The evaluation of the effectiveness of an intervention often uses clinical and experimental evaluation criterion. A clinical criterion identifies a significance. If an intervention can meet significance criterion, an individual can be considered to function normally in society (Risley, 1970). A clinical criterion is visually evaluated by setting the level of difference beforehand and then by examining whether the intervention achieves the level of difference or not.

Experimental criterion concerns the reliability of change (Risley, 1970). In single-case design, this criterion has been addressed by establishing base line, investigating difference by introducing an intervention, and replicating the base line (A-B-A design). By observing different level of performance at the end (A-B-A-B design), the repeatability can be more rigorously evaluated. In addition to visual inspection, statistical analysis can be used to evaluate the reliability of change more strongly, if applicable.

Statistical analysis can also investigate other aspects of an intervention. The statistical analysis can reveal the effect of intervention using different dependent variables, which may not show salient difference but may be more clinically important (Jones and Weinrott, 1977). This analysis is also useful when experimental criterion is equivocal (Jones and Weinrott, 1977), a base line has a trend, clinical significance is hard to obtain, a new area of research is investigated, or when intra-subject variability is high because of lack of control of experiment (Kazdin, 2012, 1975)

Statistical analysis for single-case design requires more careful attention to the statistical assumption compared to group designs. Single-case design usually has a serial dependency, which violates the critical assumption of t and ANOVA analysis. To handle such correlation problems in time, time series analysis (Jones and Weinrott, 1977) and the split-middle method of trend estimation (White, 1972) have been proposed. Time series analyses looks at level, slope, and drift. One statistical analysis for time series is using the moving average technique of time series data (Box, 2012). The split-middle method predicts future performance based on current data. However, those methods have the practical limitation of requiring sufficient data points. Another method which can avoid series dependency is randomization tests (Edgington, 1967, 1969, 1972). This method requires lengthy A-B trials. Gentile et al. suggest auto-correlation errors can be weakened by combining A phases and B phases because the auto-correlation becomes smaller if the lag between trials become large (Gentile and Klein, 1972). This method may not eradicate series dependency (Hartmann, 1974), but it requires less data points. Therefore, this method might be applicable in our study considering small number of subjects.

Researchers still express concern about the validity of statistical analysis because it sometimes does not agree with clear visual inspection results with unclear reasons (Gottman, 1973), perhaps due to a small number of data points. Therefore, This study reports both individual data and statistical analysis results in some parts.

Using single-case research and group research as a complementary tool

Because single-case research and group research can be complimentary methods (Kazdin, 2012), I utilize both in some studies. One of major concern of single-case design is the external validity. The criticism is that it might be hard to generalize the outcomes to other individuals and environments. This generalization might be achievable by showing strong and consistent effect. The applicability to other individuals can be further proven through replicated results with other samples. By conducting statistical analysis in a group level, the external validity might be addressed more thoroughly. Group research also has the inverse weak point. Even if it showed statistical significance in group, it does not necessary means that each individual may have strong effect. By investigating on both the individual-level and group-level, the external validity can be examined more thoroughly.

1.3 Outline

The remainder of this dissertation is broken into seven parts.

Chapter 2 presents a simulation study with a three-dimensional limit cycle walking model with an ankle and a foot. Ankle push-off work modulation found to be more important than we thought in restoring balance under unexpected disturbances.

Chapter 3 presents an experimental study of step-to-step modulation of push-off work with individuals with simulated amputation. Subjects reduced metabolic energy consumption, possibly by reducing their foot placement effort.

Chapter 4 presents an experimental study of step-to-step modulation of push-off work with individuals with below knee amputation. Subjects reduced intact limb control effort during stance. Forced exploration seemed to help some subjects further reduce balance-related effort and increase perceived balance assistance.

Chapter 5 presents a novel design of two degrees of freedom ankle-foot prosthesis emulator. This emulator outperformed other prostheses in controlling plantarflexion and inversion/eversion torque.

Chapter 6 presents an experimental study of the influence of ankle inversion/eversion stiffness on balance-related effort. The stiffness continuously provided a restoring torque at each step while maintaining the average torque. This actuation reduced balance-related effort.

Chapter 7 presents a simulation and experimental study of the effect of once-per-step ankle inversion/eversion torque control. Subjects reduced metabolic rate for the stabilizing controller conditions compared to the destabilizing controller conditions. This result suggests that this controller has can reduce balance-related effort.

Chapter 8 presents a summary of my approach on reducing balance-related effort for individuals with below knee amputation by altering control action at each step.

Chapter 2

The importance of ankle actuation on stability - a simulation study

Using my three-dimensional limit cycle walking model, I investigated the relative importance of ankle actuation control on stability. I compared strategies of once-per-step ankle inversion/eversion resistance control, ankle push-off work control and foot placement control. These control strategies modulated actuation parameters at each step, as a function of state deviation from nominal states at a key moment. I demonstrated that ankle push-off work control was as effective as foot placement at recovering from random variations in ground height and lateral disturbances. Ankle push-off work has control authority in not only the fore-aft direction, but also the medio-lateral direction. This authority is due to the finite distance between the location of the center of mass and the stance foot. This may explain why the ankle push-off work controller was effective at improving stability. By implementing such a controller in ankle-foot prostheses, we might help reduce balance-related effort for individuals with below knee amputation.

This work inspired human-subject experiments to study the effect of step-to-step ankle push-off work modulation, in an ankle-foot prosthesis, on balance (Chapter 3 - 4), and motivated research of alternative actuation approaches that could improve balance, specifically control of ankle inversion/eversion torque (Chapter 5, 6 and 7).

The contents of this chapter will appear in:

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A preliminary version of this work was presented at:

Kim, M., Tembulkar, T., Collins, S. H. (2014) Modulating prosthetic ankle push-off work at each step reduces balancing effort during walking. Podium presentation at World Congress of Biomechanics.

Kim, M., Collins, S. H. (2013) Ankle push-off is equally important to foot placement in stabilizing three-dimensional walking. Podium presentation at Dynamic Walking.

Kim, M.., Collins, S. H. (2012) Ankle control design to enhance sagittal stability for lower-limb amputees. Poster presentation at Dynamic Walking.

Abstract

Individuals with lower-limb amputation experience increased risk of falling, which could be partially due to a lack of active control in conventional prostheses. Inspired by stabilization strategies from planar walking robots, we hypothesized that modulating prosthetic ankle push-off could help improve amputee balance. We developed a simple three-dimensional walking model, found limit cycles at slow and normal walking speeds, and designed state-feedback controllers that made once-per-step adjustments to ankle push-off work, fore-aft and medial-lateral foot placement, and ankle roll resistance. To assess balance, we applied increasing levels of random changes in ground height or lateral impulse until the model fell down within one hundred steps. Although foot placement is known to be important to balance, we found that push-off control was more effective at recovering from both disturbances at both speeds. With push-off control, the model tolerated unexpected ground height changes of at least 7.8% of leg length and lateral impulses of at least 13 N·s, while foot placement and ankle roll resistance control tolerated disturbances of less than 1.5% and 5 N·s, respectively. Push-off work affected both fore-aft and medial-lateral motions, providing a pathway for recovery from both types of disturbance. This coupling may be especially beneficial in recovery from steps up or down, which tend to have a similar coupling. This result suggests that discrete control of ankle push-off may be more important than previously thought, and may guide the design of robotic prostheses that improve balance.

2.1 Introduction

Individuals with below knee amputation experience increased fall rates and reduced balance confidence (Miller et al., 2001a), which reduces mobility and can cause avoidance of social activity (Miller et al., 2001b). Prior research has established a connection between falling and reduced stability (Fabre et al., 2010; Major et al., 2013; Muir et al., 2010a,b) and has shown that training amputees in recovery strategies can reduce fall risk (Esquenazi and DiGiacomo, 2001; Crenshaw et al., 2013a,b). Robotic lower-limb prostheses might also prevent falls by improving stability during walking, although most development efforts to date have been focused on other aspects of gait such as average joint kinematics or overall energy use (Herr and Grabowski, 2012; Goldfarb et al., 2013; Hitt et al., 2009; Caputo and Collins, 2014b). Stability-related outcomes have been compared across devices in some cases (Aldridge et al., 2012; Gates et al., 2013b; Lawson et al., 2013), but results thus far have been inconclusive.

The best understood methods for stabilizing gait involve control of foot placement and center of pressure, but these are difficult to implement in robotic ankle-foot prostheses. Simple models of walking suggest that foot placement is an efficient approach to balance, since small adjustments prior to heel strike can have large effects on the trajectory of the ensuing step (Kuo, 1999). This phenomenon is central to 'capture point control', used for disturbance recovery in humanoid robots (Pratt et al., 2006). Experimental studies with able bodied subjects (O'Connor et al., 2012; Collins and Jackson, 2013) and individuals with below knee amputation (Gates et al., 2013b) suggest that humans use a similar approach during walking. In humanoid robotics, control of the center of pressure (often referred to as the 'zero moment point') between the foot and the ground has also been central to many stable walking algorithms (Kim et al., 2007). Humans also seem to modulate center of pressure location for balance to some degree (Hof et al., 2010), and individuals with above knee amputations seem to exhibit increased reliance on this strategy in the intact limb (Hof et al., 2007). These two control approaches are strongly linked; foot placement constrains the region of possible center of pressure locations and defines the location corresponding to zero ankle torque, while center of pressure adjustment through ankle activity is akin to slightly moving the foot after contact has been established. Although these forms of control can be effective and seem to be commonly used for balance by humans, they would be difficult to implement with a robotic ankle prosthesis. Foot placement control is most easily achieved through hip actuation, while center of pressure control is most effective using a wide, flat foot with multiple actuated degrees of freedom, neither of which are currently available in lower-limb prostheses.

Ankle push-off work modulation is a promising alternative stabilization method. Regulating system energy is necessary for stable locomotion, and simple two-dimensional models of gait show that system energy can be strongly affected by the magnitude of work produced by active plantarflexion of the trailing ankle during transitions between steps (Ruina et al., 2005). Modulation of this ankle 'push-off', in concert with control of foot placement, has been used to stabilize simple two-dimensional walking robots (Bhounsule et al., 2012; Hobbelen and Wisse, 2008). Three-dimensional walking seems to be less stable, however, with the least stable modes corresponding to medial-lateral motions (Kuo, 1999). Push-off work modulation might still be effective in such systems if ankle push-off were to have some control authority over medial-lateral motion of the body. If ankle push-off control were found to be effective at stabilizing three-dimensional walking, it would help explain balance deficits in individuals with amputation below the hip of the effected limb. It would also be feasible to implement push-off modulation in active ankle-foot prostheses, which could improve balance for millions of individuals with lower-limb amputation.

Simulations of limit cycle walking could provide well-controlled comparisons of the effectiveness of push-off work, foot placement and center of pressure control techniques. Limit cycle models can capture features of the basic dynamics of human gait while remaining simple enough to be intellectually accessible. Such models seem to help explain, for example, how step length relates to energy use (Kuo et al., 2005) or why crouch gait is typically accompanied by stiff-knee gait (van der Krogt et al., 2010). Limit cycle models are especially useful for the study of stability, where they allow a level of precision and control that can be difficult to achieve experimentally. They have previously been used to illustrate the utility of active foot placement as a means of stabilizing three-dimensional walking (Kuo, 1999), with results that are qualitatively consistent with those from experiments in humans (Donelan et al., 2004; O'Connor et al., 2012; Collins and Jackson, 2013). Limit cycle models have also been used to design push-off work controllers for two-dimensional walking robots (Hobbelen and Wisse, 2008), resulting in a machine that set the distance record for legged robots (Bhounsule et al., 2012). A comparison of these control techniques with three-dimensional models of gait is therefore likely to provide useful qualitative insights into their strengths and weaknesses, and could lead to the design of improved prosthesis controllers.

The most meaningful measure of stability in this context seems to be the maximum random disturbance that can be tolerated without falling. Many other candidate metrics exist, but do not seem well correlated with the likelihood of falling under real-world conditions (Bruijn et al., 2013). Maximum floquet multipliers are easily obtained by linearizing a dynamic system around a fixed point, but moderate disturbances often move the system outside the linear region for which they are relevant. Basins of attraction capture behavior in full nonlinear regions, but do not include information about which directions in state space are likely to be encountered, making interpretation difficult. Gait sensitivity norms (Hobbelen and Wisse, 2007) measure a combination of convergence rate and performance during convergence, but rely on a gait indicator that must be calibrated against a more meaningful measure of stability. Maximum allowable disturbance approaches have none of these issues; they include nonlinear behavior, implicitly capture the relevance of state error direction, and need not be calibrated against additional measures. Maximum allowable disturbance is calculated by selecting a disturbance relevant to real-world falls, such as ground irregularity (Song et al., 2013) or lateral pushes (Donelan et al., 2004), and gradually increasing the magnitude of the disturbance until the system can no longer recover. A disturbance should be applied on every step so as to penalize solutions that recover slowly and are therefore susceptible to multiple consecutive disturbances. This means many walking steps must be simulated to evaluate each controller. Simulating more steps increases accuracy but also increases computational cost, and so a minimum acceptable number of steps must be chosen carefully.

Walking speed can also affect stability, and might interact with disturbance recovery strategy. Walking speed is correlated to changes in gait pattern (Krasovsky et al., 2014), fall risk (Espy et al., 2010), and ability to recover from some types of disturbances (Kadono and Pavol, 2013). Considering different disturbances at different walking speeds would therefore lend insight into the conditions under which one or another recovery method is likely to be most effective.

In addition to maximum disturbance rejection, the energy required to balance at sub-maximal disturbance levels can differentiate control strategies. Active balance during walking seems to require the expenditure of meaningful amounts of metabolic energy in humans (Donelan et al., 2004; IJmker et al., 2013), which increases in the presence of sensory manipulation or ground height disturbances (O'Connor et al., 2012; Voloshina et al., 2013). Qualitative differences in energy requirements across control strategies could also be explored in simulation.

Simple, low-order control strategies are preferable when transferring strategies from simulation to hardware. These tend to offer increased robustness against errors in the model of the human and to rely on less sensor information. In simulation, full state linear feedback control, e.g. derived as a linear quadratic regulator (LQR), is likely to result in effective disturbance rejection. Performance outside the linear region can be further improved using optimization of the gain matrix, for example using a covariance matrix adaptation evolutionary strategy (CMA-ES). A simulation model of human walking will necessarily be much simpler than the real system, however, and so only the most basic control strategies are likely to have relevance. It is also difficult for the prosthesis to directly measure most of the human's activity, making it desirable to develop controllers based only on local state information. Often, full state feedback control can be approximated by a lower-order controller with only small reductions in performance (Sala and Esparza, 2003). One way to tune candidate lower-order controllers is to train a neural network to produce the same results as the original controller (Zurada, 1994).

This simulation study was designed to compare the effectiveness of ankle push-off control against foot placement and ankle inversion/eversion control in three-dimensional walking. We hypothesized that ankle push-off control could result in similar maximum tolerable disturbances and energy consumption as these more widely used strategies, while relying only upon actuation available to a prosthetic ankle. We also hypothesized that it would be possible to derive a simple, robust form of the ankle push-off controller suitable for use in hardware experiments.

We explored some of these ideas in a preliminary study using a simpler model and standard LQR control, which was presented at ICORR in 2013 (Kim and Collins, 2013). In the present study, we make comparisons of more complete models, which walk at multiple speeds under multiple disturbances. We develop more complete controllers that have been optimized for disturbance tolerance using CMA-ES. We also investigate the effects of control type on energy use, and develop reduced-order controllers suitable for implementation in hardware.

2.2 Methods

We developed a three-dimensional limit cycle walking model with hip and ankle actuation and used it to compare the capacity of foot placement, ankle inversion/eversion control, and push-off work control to stabilize gait against ground height disturbances and lateral impulse disturbances. The model has finite pelvis width, two straight legs attached to the pelvis via hip joints, and massless feet connected to the legs via ankle joints. Hip joints were controlled to modulate step length and step width, while ankle joints were controlled to change ankle roll resistance and ankle push-off work. After developing nominal controllers for the hip and ankle joints, we designed discrete stabilizing controllers that modulated step-length, step-width, ankle roll resistance and ankle push-off work once per step. We compared the performance of each controller in terms of maximum tolerable random disturbances in ground height and lateral impulse. Finally, we developed a hardware-implementable version of the ankle push-off controller and compared its performance to full state feedback.

2.2.1 Model

Mathmatical model description

We developed a model with a pelvis, two straight legs, and two feet. The pelvis and legs were connected via hip joints that allowed continuous flexion-extension and once-per-step changes in adduction-abduction angle (Fig. 2.1, as in (Kuo, 1999)). The legs and feet were connected via ankle joints that allowed plantarflexion-dorsiflexion and inversion/eversion. The feet and ground were connected either rigidly, by a toe pitch joint, or by both a toe pitch joint and a toe yaw joint, depending on phase of stance.

Mechanical parameters of the model were based on human anthropometrics (Winter, 1991; Chandler et al., 1975). Hip width was 0.3 m and leg length was 1 m.

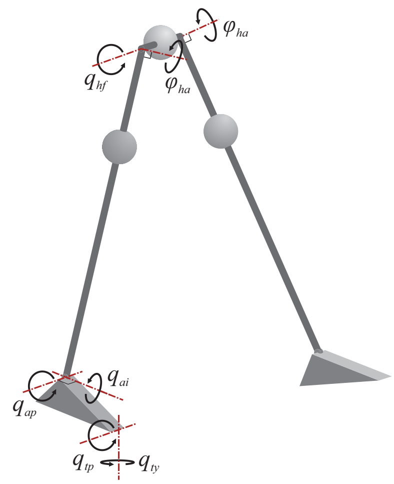


Figure 2.1: Model schematic. The model had a finite-width pelvis, two straight legs, and two massless feet. The hip had a flexion-extension joint (q_{hf}), and an abduction angle (ϕ_{ha}) that could be changed once per step at mid-stance. The ankle of the stance leg had a plantarflexion joint (q_{ap}) and an inversion (roll) joint (q_{ai}). The stance foot was connected to the ground either rigidly, through a toe pitch joint (q_{tp}), or through both a toe pitch and a toe yaw joint (q_{ty}), depending on phase of the gait cycle. All degrees of freedom are defined such that, beginning at the ground, positive rotation causes the subsequent segment to move in the direction indicated by the arrow.

Foot length from heel to toe was 0.25 m, while the horizontal distance from ankle to toe was 0.19 m, foot height from base to ankle was 0.09 m, and foot width (used to check center of pressure feasibility) was 0.1 m. Nominal step width, set by choice of nominal hip abduction angle, was 0.15 m. The pelvis had a mass of 54 kg, located at its center, and a rotational inertia of 10 Kg·m^2, which together approximated the mass properties of the head, arms and torso. Each leg had a mass of 10 kg, with center of mass located 0.3 m from the hip joint. The feet were treated as massless.

Dynamics

During each walking step, the model went through a double support phase, a fully actuated single support phase, and, on most steps, an under-actuated single support phase. During double support (Fig. 2.2) the leading foot was rigidly attached to the ground while the trailing toe was connected to the ground through a two degree of freedom joint that allowed both pitch and yaw rotations. The yaw degree of freedom

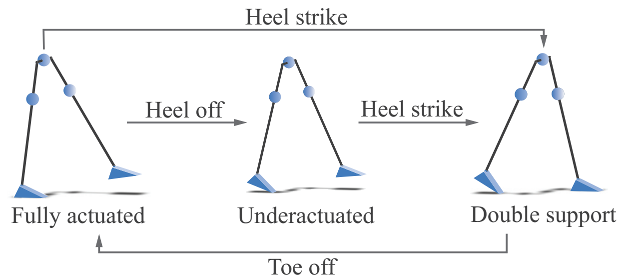


Figure 2.2: Model gait phases. During a walking step, the model went through at least two of three possible phases: fully actuated single support, underactuated single support, and double support. From the fully actuated phase, the model could transition to either double support, if foot strike was detected, or to underactuated single support, if stance heel rise was detected. From the underactuated phase, the model transitioned to double support when the swing foot touched the ground. From the double support phase, the model transitioned to fully actuated single support when the ground reaction force at the toe of the stance foot became zero.

gave the closed-loop kinematic chain two degrees of freedom, resulting in more natural motions during double support. Toe off occurred when the vertical component of the reaction force of the trailing toe went to zero, leading to single support. During the initial portion of single support, the stance foot was fixed to the ground, allowing full actuation of the resulting three degrees of freedom (two at the stance ankle and one at the hip). Heel rise occurred when the vertical component of the reaction force of the heel of the stance foot went to zero, leading to the under-actuated phase. During the under-actuated phase of single support, the foot was connected to the ground through a hinge joint that allowed pitch rotation, with four degrees of freedom in total. Foot strike was detected when the base of the swing foot reached ground height, after which the model underwent a perfectly inelastic collision and transitioned into double support. On most steps foot strike occurred during the under-actuated phase of single support, but with large disturbances foot strike sometimes occurred during the fully actuated phase of single support.

Equations of motion for each phase were obtained using the Dynamics Workbench (Kuo, 2012), a software program based on Kane's method. State trajectories for each step were calculated using forward numerical integration. The heel strike collision was modeled using an impulse-momentum approach, in which post-collision velocities were obtained as a function of pre-collision states. We modeled the body as an open kinematic chain during this collision, and solved for the impulse on the leading foot that would cause it to have zero velocity following the collision. We simultaneously solved for the post-collision velocities of the trailing toe pitch and yaw joints, ankle plantarflexion and inversion/eversion joints, and hip flexion joint by performing an angular momentum balance about each joint that included the effect of the impulse on the leading foot.

Limit cycles were found using a gradient search algorithm that altered initial conditions to minimize error between the initial and final states of a walking step. Limit cycles were found at two human-like speeds and step lengths, 1.00 $\mathrm{m{\cdot}s^{-1}}$ with 0.63 m steps and 1.25 $\mathrm{m{\cdot}s^{-1}}$ with 0.70 m steps, approximating the range of preferred speeds and step lengths of high-activity individuals with lower-limb amputation (Hsu et al., 2006). Limit cycles with desired characteristics were found using a nested gradient search approach that altered nominal control parameters to minimize error between desired and observed speed and step length (Collins et al., 2009).

2.2.2 Actuation and control

Hip and ankle joints were controlled in two layers: a continuous low-level controller achieved target values for step length, step width, ankle roll resistance and ankle push-off, while a discrete high-level controller set these targets once per step (Fig. 2.3).

Low-level, within-step control

Hip flexion-extension torque was continuously controlled to achieve desired step length. We used proportional-derivative control of hip flexion angle, where the set point was ϕ_{hf} and the nominal value corresponded to the preferred step length for humans. We chose relatively high stiffness and damping gains, such that the hip

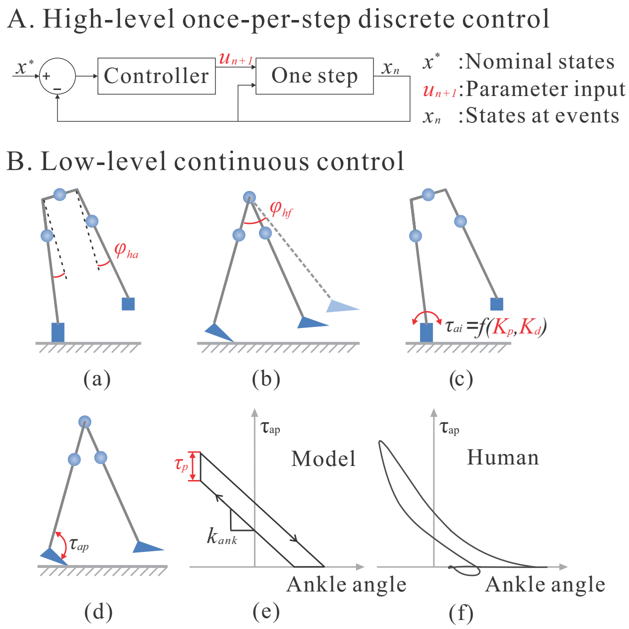


Figure 2.3: Control architecture. Control was performed in two layers: **A** high-level, discrete control that used linear state feedback to make adjustments to low-level parameters once per step, and **B** low-level control that continually regulated joint torques in accordance with parameters during the course of a step. The actuation parameters used in low-level control were: (a) hip abduction angle, ϕ_{ha}, a fixed parameter only changed at mid-stance, which affected step width, (b) target hip flexion angle, ϕ_{hf}, the set point in a proportional-derivative controller on hip flexion torque, which affected step length, (c) ankle inversion/eversion stiffness, K_p, and damping, K_d, gains in a proportional-derivative controller on ankle inversion/eversion torque, which affected roll resistance and medial-lateral center of pressure location, and (d-e) ankle plantarflexion torque offset, τ_p, an offset in ankle torque during the phase when joint velocity was positive, which affected ankle push-off work. (f) Default values of torque offset and ankle stiffness were chosen to approximate the torque-angle curve observed for humans (Caputo and Collins, 2014b).

flexion controller settled at target step length within 90% of the stance period at the limit cycle. This resulted in improved fore-aft stability (Wisse et al., 2005).

Hip abduction-adduction angles were set once per step to achieve desired step width. We discretely changed the rigid hip abduction angle paramter, ϕ_{ha}, at mid-stance in the manner of (Kuo, 1999). The nominal value of ϕ_{ha} corresponded to the preferred step width for human walking.

Ankle inversion/eversion torque was continuously controlled to provide desired levels of resistance. Inversion/eversion torques followed a proportional-derivative control law, with gains of K_p and K_d and set point angle and angular velocity of

θ_0 and $\dot{\theta}_0$. The nominal values for K_p and K_d were both zero, and the nominal values of θ_0 and $\dot{\theta}_0$ corresponded to the value of eversion angle and angular velocity just after heel strike during limit cycle motions.

Ankle plantarflexion torque was continuously controlled to provide desired levels of ankle push-off work. Torque was applied as a function of ankle angle and direction of motion, as depicted in Fig. 2.3(e), or:

$$\tau = -k_{ank}(\theta - \theta_0) + max(0, sign(\dot{\theta})) \cdot \tau_p \tag{2.1}$$

where τ is ankle plantarflexion torque, k_{ank} is ankle stiffness, θ is ankle plantarflexion angle, θ_0 is nominal ankle angle, $\dot{\theta}$ is ankle angular velocity, and τ_p is the plantarflexion torque offset. The values of k_{ank} and θ_0 were selected so as to approximate the average torque-angle curve of the human ankle, while the nominal value of τ_p was set during the search for a limit cycle with desired speed and step length. The curve formed by this function in angle-torque space is a work loop, with the area inside corresponding to net ankle work during a step. Because peak dorsiflexion angle is relatively consistent, τ_p is approximately proportional to net ankle work.

High-level, once-per-step control

We developed several high-level controllers that altered target values of step length, step width, ankle roll resistance, ankle push-off work, or combinations of these low-level control parameters once per step in order to maintain balance. The system was discretized by sampling states once per step at a predefined state event, or a Poincaré section. Each high-level controller was discrete and linear, having the form:

$$u_{n+1} = u^* - K(x_n - x^*) \tag{2.2}$$

where u_{n+1} is a vector of control inputs (some combination of ϕ_{hf}, ϕ_{ha}, K_p, K_d or τ_p) for the $n{+}1^{th}$ step, u^* is the nominal vector of control inputs corresponding to limit cycle motion, K is the gain matrix of the discrete linear controller, x_n is the state vector at the end of the n^{th} step, and x^* is the state vector corresponding to limit cycle motions. For most high-level controllers, we used full state feedback, consisting of the angles and angular velocities of all model joints.

High-level control decisions were made at mid-stance for step length and step width control, and at the instant following heel strike for ankle roll and push-off control. At mid-stance, velocities and displacements of the center of mass were well captured, while sufficient time remained to place the swing foot (Bhounsule et al., 2012). At heel strike, the time delay between control decision and control action (in either trailing ankle push-off or leading ankle roll torque) was minimized.

We developed discrete linear approximations of the dynamics of the model and control inputs and used these to generate feedback gain matrices with a linear quadratic regulator approach. We approximated the discrete dynamics as $x_{n+1} = A \cdot x_n + B \cdot u_n$, where x_{n+1} is the state at the end of the $n{+}1^{th}$ step, A is the state transition matrix, x_n is the state at the end of the n^{th} step, B is the control input matrix, and u_n is the control input on step n. We used a finite differencing approach about the fixed point to obtain the A matrix and to obtain B matrices corresponding to each set of control inputs. These models were then used to generate linear quadratic regulators (LQR), each consisting of a gain matrix, K, for use in Eq. 2.2.

We found that disturbance rejection could be significantly improved by refining the gain matrix using a genetic algorithm. This improvement likely relates to the fact that the LQR result is only optimal in a narrow linear region, and does not utilize information about the types of disturbances likely to be encountered by the system. The value of K determined using LQR was used as an initial seed in a

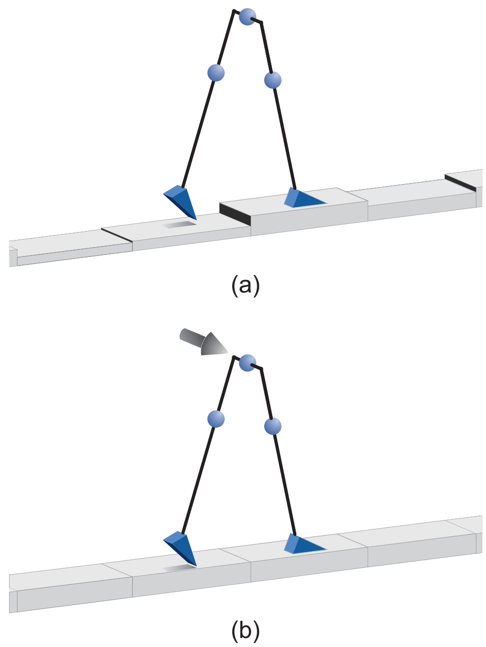


Figure 2.4: Ground height disturbance (a) and lateral impulse disturbance (b). The ground was modeled as a series of flat surfaces, each centered below the landing foot, and each with a randomly chosen height with respect to a constant reference. Possible tripping of the swing foot was not considered. The magnitude of the disturbance was defined as the maximum possible change in height between two consecutive steps. Similarly, lateral disturbance was applied at the beginning of each step.

covariance matrix adaptation evolutionary strategy (CMA-ES, (Hansen, 2006)). We used the cost function $f = 1/h$, where h was the maximum tolerable ground height disturbance for the full non-linear system. We used a population size of 30 to optimize computation time. The algorithm typically underwent about 150 to 300 generations before convergence. The resulting optimized gain matrices, K, were used in across-controller comparisons.

2.2.3 Stability Measure

We quantified stability as the maximum random floor height disturbance and the maximum random lateral impulse disturbance that the model could tolerate for one hundred steps without falling. Before each walking bout, a bounded random array of floor heights and impulses (Fig. 2.4) were generated. The magnitude of the floor

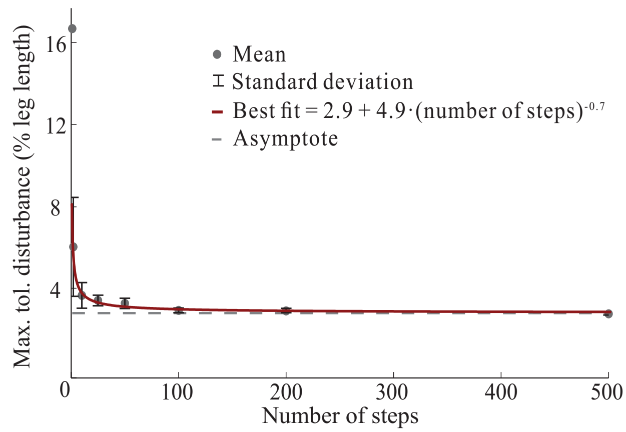


Figure 2.5: Maximum tolerable ground height disturbance versus number of steps tested. Dots and whiskers are the means and standard deviations of five tests of maximum allowable disturbance using different random ground patterns. We fit data with an exponential curve, shown in red. The mean approached a constant as the number of steps increased, shown as a dashed line, while standard deviation approached zero. At 100 steps, the mean maximum tolerable disturbance value was within 2% of the asymptote.

height disturbance was defined as the difference between the upper and lower bounds of possible heights. Maximum tolerable disturbance was found by slowly increasing the magnitude until the model was unable to complete a predefined number of steps without falling.

To determine an appropriate number of steps, we tried several values and compared disturbance tolerance. We generated five sets each of random height distributions having lengths from 1 to 500 steps, and calculated the mean and standard deviation of maximum tolerable disturbance at each length (Fig. 2.5). We found that maximum tolerable disturbance appeared to converge to within 2% of the final value when at least 100 walking steps were tested, and that the standard deviation across different randomly-generated ground patterns was also less than 2% for this number of steps. We therefore used 100 continuous steps in tests of disturbance tolerance.

2.2.4 Energy expenditure measure

We used positive mechanical work performed by hip and ankle joints to quantify energy use. Since this system is periodic and does not, on average, change speed or height, total negative and positive mechanical work are equal and opposite on average. We calculated energy use for sub-maximal disturbance levels, ranging from no disturbance to maximum tolerable disturbance, in order to capture changes in energy consumption associated with balance.

2.2.5 Hardware-implementable control

For the most effective full-state feedback controllers, we developed reduced-order versions suitable for implementation in robotic prosthesis hardware. Sensory information was limited to local measurements only, including step period and ankle joint angles and velocities. We first used linear regression to calculate new gain matrices that used reduced sensory information to reproduce the full-state feedback control inputs with least squared error. These gains were refined for the non-linear system using a genetic algorithm (CMA-ES). In parallel, we used a neural network approach to design candidate nonlinear controllers, first using full-state control inputs as training data and later refining node gains using CMA-ES.

2.2.6 Simulation experiment

We compared disturbance tolerance among the most effective high-level controllers. In particular, we compared controllers based on step length and step width (foot placement), ankle roll resistance (both stiffness and damping), ankle push-off work, and the combination of all five control inputs. For a representative model of gait (a model with normal walking speed), we compared energy use across controllers, and compared full-state feedback controllers with their reduced-order hardware-implementable analogues.

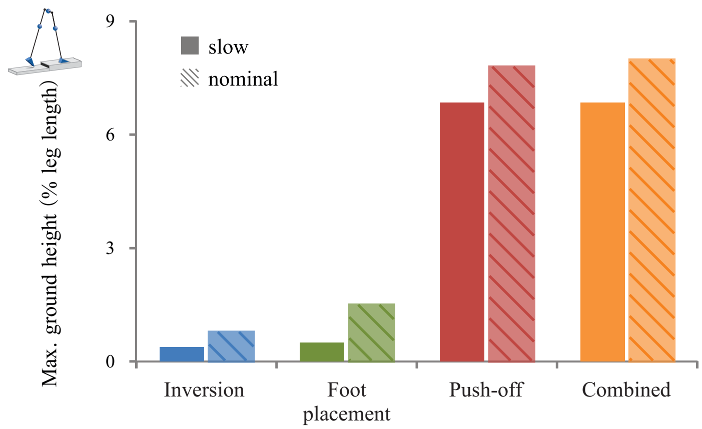
2.3 Results

Once per step control of ankle push-off work resulted in better disturbance rejection than control of foot placement or ankle roll resistance for both disturbance types and gait speeds. Performance with push-off work control alone was nearly as effective as controlling all inputs together. Push-off work modulation allowed the model to withstand random changes in ground height of up to 7.8% of leg length (0.085 m) compared to 1.5% leg length (0.016 m) with foot placement, or about five times greater disturbance tolerance, at normal walking speed (Fig. 2.6). The push-off controller tolerated random lateral disturbances of up to 13 N·s, compared to 5 N·s with foot placement. For larger ground height disturbances, the push-off controller failed to achieve ground clearance with the swing foot in at least one step in one hundred. For step width, foot placement and ankle roll resistance controllers, the model fell sideways with higher disturbances, consistent with prior modeling results (Kuo, 1999). See the Appendix for a complete table of maximum tolerable disturbance values.

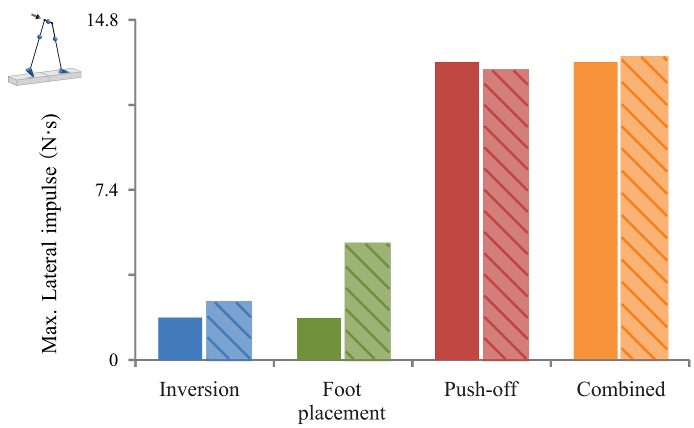
Optimization of gain matrices using a genetic algorithm improved disturbance tolerance for all controllers, but did not significantly affect the trend across controllers. For example, maximum tolerable disturbances using the unmodified gain matrices derived with LQR were 2.9% leg length (0.032 m) and 1.0% leg length (0.0102 m) using push-off work control and step width control, respectively, at the normal walking speed. See the Appendix for a complete set of gain matrices.

Other candidate measures of stability, including maximum floquet multiplier and gait sensitivity norm, did not correlate well with maximum tolerable ground height disturbance. See the Appendix for a complete table of stability metrics.

In the linearized system model, ankle push-off had strong control authority over



(a)



(b)

Figure 2.6: Maximum tolerable disturbance in ground height (a) and maximum tolerable lateral impulse disturbance (b) versus high-level control approach. Bars represent the maximum, bounded, random, ground-height variation (a) and lateral disturbance variation (b) that the model could tolerate for 100 steps without falling. Solid bars are for slow walking (1.0 m·s^{-1}) and patterned bars are for a normal walking speed (1.25 m·s^{-1}). Four different high-level controllers were tested: Foot placement, based on ϕ_{ha} and ϕ_{hf}; Ankle roll resistance, based on K_p and K_d; Ankle push-off work, based on τ_p; and Combined inputs, based on ϕ_{ha}, ϕ_{hf}, K_p, K_d and τ_p. Ankle push-off work control led to the greatest disturbance tolerance among individual methods.

Table 2.1: Control authority unit vectors

	Control input				
State	ϕ_{hf}	ϕ_{ha}	Kp	Kd	τ_p
q_{tp}	-0.050	-0.041	-0.052	-0.029	0.117
q_{ap}	-0.003	0.037	0.003	0.000	-0.152
q_{ai}	-0.005	-0.048	0.247	0.151	0.119
q_{hf}	0.088	-0.008	-0.026	-0.017	0.082
$\dot{q}_{tp}$	-0.679	-0.472	0.028	0.179	-0.620
$\dot{q}_{ap}$	0.718	0.594	-0.228	-0.343	0.396
$\dot{q}_{ai}$	-0.037	-0.643	0.939	0.909	0.634
$\dot{q}_{hf}$	-0.107	-0.076	-0.021	0.011	0.001

both medial-lateral and fore-aft motions. The column-wise normalized control input matrix (Table 2.1) illustrates the relative influence of each control input on each model state. In the ankle push-off column (τ_p), the largest value was associated with ankle eversion velocity, $\dot{q}_{ai}$; as push-off work increased, eversion velocity at the end of the subsequent step became more positive. That is, pushing off more caused the model to roll outwards more during the ensuing step and to be rolling back towards the swing foot less at the time of the next heel strike. The next largest values indicate that as push-off work increased, trailing toe angular velocity ($\dot{q}_{tp}$) became more negative and trailing ankle angular velocity ($\dot{q}_{ap}$) became more positive, though with half the effect size. That is, pushing off more caused the vertical and forward components of the center of mass velocity to be increased at the end of the subsequent step. Other control inputs had either little effect on vertical and fore-aft motions (K_p and K_d) or opposite coupling of effects (ϕ_{hf} and ϕ_{ha}).

Energy use increased for large ground height disturbances due to an increase in walking speed and associated collision dissipation. For all high-level controllers, changes in positive joint work were negligible for low levels of disturbance (below 1% changes in ground height, Fig. 2.7). In this region, step-by-step differences in energy use due to control actions tended to cancel out over many steps. At higher levels of disturbance, only ankle push-off work control was able to maintain balance. As

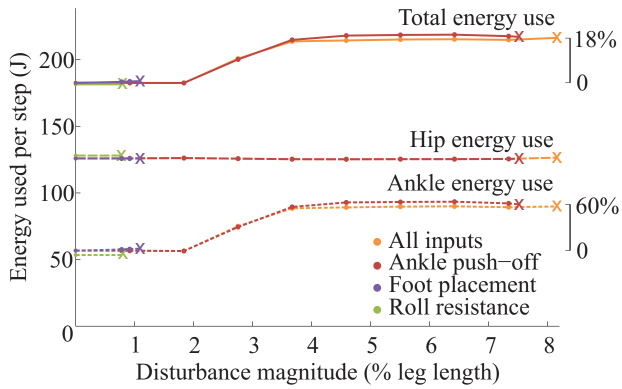


Figure 2.7: Energy expenditure of the nominal speed model under the ground height disturbances as a function of disturbance magnitude. Solid lines represent total energy use, long-dashed lines the component used at the hip and short-dashed lines the component used by the ankle. Colors represent different high-level controllers, with X's indicating the point at which the model could no longer tolerate disturbances. Energy used at the hip was unchanged with increasing disturbance. Ankle energy use increased in the region between about 2% and 4% of leg length, corresponding with an increase in walking speed and a shift to a gait pattern in which the model tended to transition into double support without heel rise in the stance foot.

disturbance magnitude increased from around 2% to 4% leg length, walking speed increased by about 20%, from 1.25 m·s^{-1} to 1.54 m·s^{-1}. This change in speed arose through dynamic interactions between the disturbance, resulting state errors, the optimized gain matrix, and resulting ankle push-off work. At higher speeds, trailing ankle stiffness, k_{ank}, was too low to cause the stance heel to rise prior to leading leg collision. This led to a sharp increase in the prevalence of steps in which heel rise did not occur prior to heel strike, from 0% of steps around 2% leg length disturbances to more than 90% of steps around 4% leg length disturbances. With the stance foot flat on the ground prior to heel strike, the center of mass velocity was directed more downwards, leading to greater energy dissipation in the ensuing collision. Over the same range, overall energy use increased by about 20%, which was entirely accounted for by a 60% increase in positive mechanical work at the ankle joint.

We explored many reduced-order control strategies, and found that reasonable performance could be achieved using push-off work control based on sensed ankle inversion/eversion velocity alone. Using linear regression, target push-off work matched that calculated using full-state feedback with less than 1% error. After optimization of the gain matrix (in this case $K \in \mathbb{R}^1$) using a genetic algorithm, the reduced-order push-off work controller tolerated ground height disturbances of 1.8% leg length. This reduced-order feedback law essentially stated that when the model was rocking back towards the swing foot side too slowly at heel strike, it should push off harder with the stance foot. More precisely, the controller commanded push-off work in proportion to the difference between the measured and expected ankle inversion velocity, with positive velocity defined as causing the model to move in the direction of the stance leg:

$$\tau_p = K \cdot (\dot{q}_{ai} - \dot{q}_{ai}^*) \tag{2.3}$$

where τ_p is the torque offset (related to net ankle work), K is a positive scalar, $\dot{q}_{ai}$ is measured ankle eversion velocity (related to medial-lateral center of mass velocity), and $\dot{q}_{ai}^*$ is ankle eversion velocity at the fixed point of the limit cycle.

2.4 Discussion

We compared the effectiveness of once-per-step control of ankle push-off work, foot placement, and ankle roll resistance at recovering from random disturbances in ground height and lateral impulse. Control of push-off work was by far the most effective approach, tolerating changes in ground height and lateral impulse that were at least two times greater than any other strategy, regardless of the speed of the model. This strongly suggests that ankle push-off work can be an important contributor to balance maintenance in the presence of the types of disturbances expected in human environments.

Although most explanations of the role of ankle push-off have focused on effects in the sagittal plane, ankle push-off also has strong control authority over medial-

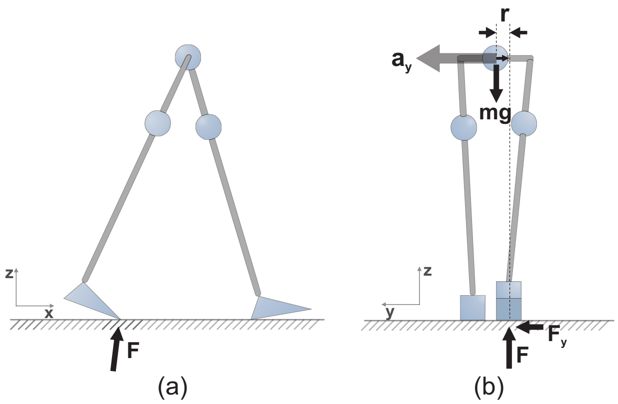


Figure 2.8: Push-off affects frontal plane dynamics. The force generated by push-off ($\mathbf{F}$), usually described in (a) the sagittal plane, can also affect (b) frontal plane motions. With finite medial-lateral distance between the foot and the center of mass ($\mathbf{r}$), the combined effects of push-off force and body weight lead to a medial-lateral force at the foot, $\mathbf{F}_y$, and a medial-lateral component of body acceleration ($\mathbf{a}_y$). If one neglects rotational inertia about the center of mass, and three-dimensional coupling, lateral acceleration is proportional to push-off force as $\mathbf{a}_y = \frac{1}{m} \cdot \frac{\mathbf{r}}{L} \cdot \mathbf{F}$, where L is leg length.

lateral motions, the coupling of which may have been advantageous for recovery from ground height disturbances. Under typical conditions, push-off torques lead to both vertial and medial-lateral components of force at the trailing toe (Fig. 2.8), thereby contributing to side-to-side accelerations of the center of mass. Increased ankle push-off caused reduced roll velocity towards the swing foot and increased forward and vertical velocity of the center of mass at the time of the next heel strike. This combination of effects may be well tuned to counteract unexpected changes in ground height. A step down tends to add kinetic energy to the system, with increases in both the fore-aft and medial-lateral components of velocity. During the ensuing step, without intervention, the model tends to roll outwards more, then roll back less, and to have increased forward velocity. Pushing off less at the beginning of such a step helps remedy both problems. A complementary set of events occurs when the model steps up unexpectedly. None of the other control inputs had this combination of effects, which may help explain their lower capacity to recover from this type of disturbance.

The most effective of the reduced-order push-off controllers further illustrates the mechanism by which push-off modulation helped the system to tolerate ground disturbances. In essence, the controller commanded more push-off when the model was rocking back towards the swing foot side too slowly at the moment of heel strike. This tends to happen when the swing foot contacts the ground earlier than normal, which tends to happen during unexpected steps up. Under such circumstances, pushing off more helps not only with medial-lateral motions but with fore-aft motions as well, since increased energy input is required to vault over the stance leg in the sagittal plane. A similar set of tendencies is true for steps down. In the context of random disturbances in ground height, this reduced-order control law could be summarized as "if you step in a hole, push off less; if you step on a bump, push off more". What is surprising is that the most important effects of such pushes are on frontal plane dynamics.

The effect of push-off work modulation on frontal plane dynamics appears to be more influential than was previously thought. When the models were exposed to pure lateral disturbances, the best balance restoring ability was shown by the push-off work controller among the individual control methods. This favorable results may have occurred because in three-dimensional walkings, the lateral disturbance can change forward as well as lateral velocity. These changes can also affect collision work. By modulating push-off work at the end of the stance phase of the same step, such changes can be regulated, which explains why the model outperformed other methods for lateral disturbance.

Discrete ankle push-off control resulted in the greatest disturbance tolerance for all control design approaches and nominal gait variations that we explored. In addition to low-level hip flexion control with high-gain feedback, we also tested models with spring-like hip flexion (see Appendix) similar to those used in prior simple dynamic models (Kuo, 2001). Spring-like hip flexion control resulted in reduced disturbance tolerance in ground height for all high-level controllers (e.g. 0.66% leg length with push-off control) but the same trend across controllers (e.g. 0.13% leg length with step width control). We also examined limit cycles with larger nominal step width (see Appendix). This resulted in a slight increase in both energy use and maximum tolerable disturbance, as reported in previous studies (Kuo, 1999), but did not affect the relative disturbance tolerance of high-level controllers. The effectiveness of ankle push-off control across low-level control strategies and control design methods suggests that the approach could be robust enough to apply to human walking.

In this model, there was no increase in energy cost associated with step-by-step control actions to maintain balance. For small disturbances, no change in energy use occurred for any control type. For larger disturbances, push-off control actions tended to increase walking speed, which led to increased nominal energy cost. In optimizing the controller, rejection of errors in the more fall-prone medial-lateral direction might have been achieved at the cost of poorer rejection of errors in the fore-aft direction. Increased walking speed might also have been a strategy for improving nominal stability discovered by the genetic algorithm. Whatever the cause, increased energy use at high disturbance levels was not due to step-by-step changes in joint work associated with balance. This bodes well for the application of push-off control in robotic prostheses, since it might not require an increase in average power output.

Although we only considered linear state feedback controllers, the results reported here are consistent with prior simulation studies utilizing nonlinear control of foot placement or center of pressure. In general, nonlinear control encompasses a larger design space and might be expected to result in better performance. It is possible that disturbance tolerance could be improved more with nonlinear control for approaches using foot placement and center of pressure than those using ankle push-off. To provide context, we repeated tests applied in two previous simulation studies examining stability and compared outcomes (see Appendix for details). We found that the best foot placement controller derived here tolerated similar maximum downhill slopes as a prior foot placement approach (-2.5° compared to -3° slope in (Wang et al., 2009)). Similarly, the ankle inversion/eversion resistance controller derived here tolerated similar steps down as a prior center of pressure approach (0.057 m compared to 0.025 m in (Kajita and Tan, 1991)). These comparisons suggest that the linear controllers used here do not put foot placement or center of pressure techniques at a disadvantage. Still, with the addition of techniques such as LQR trees (Tedrake et al., 2010), we would expect improvements in disturbance tolerance for all control inputs. A more complete model might also have lent insights into the effectiveness of other balance strategies, such as those using the arms and torso. Both the trunk (Benallegue et al., 2013) and arms (Ortega et al., 2008; Bruijn et al., 2010) have been suggested as contributors to stability in human gait, and these should be investigated in future studies.

The finding that ankle push-off work was more effective than foot placement and center of pressure control might be specific to random ground height and lateral impulse disturbances. The coupling of effects on fore-aft and medial-lateral center of mass velocity through ankle push-off appears to have been beneficial in these cases, and different couplings might be advantageous with, e.g., disturbances in the form of constant (cross) slopes. We also found that random disturbances resulted in different relative effectiveness of control inputs than, e.g., a single step down followed by unperturbed walking. In the single step down case, both ankle push-off and foot placement resulted in the same maximum tolerable disturbance (8.3 %leg length, 0.090 m), with the limiting factor being foot clearance on the next step.

2.5 Conclusions

In this study, we have shown that once-per-step control of ankle push-off work can be more effective than foot placement and center of pressure control at recovery from random fluctuations in ground height and lateral impulse. The key to this result seems to have been that push-off provides a useful combination of effects on both medial-lateral and fore-aft motions. We designed an effective reduced-order controller based only on ankle inversion/eversion velocity, which commanded more push-off work when the model was rolling back towards the swing leg more slowly than expected. This reduced-order controller may be relatively easy to implement in a robotic ankle prosthesis, since it requires only local state information and has only one gain parameter that might require tuning. The technique also requires no additional power on average, meaning that its incorporation would not require larger batteries. We are currently testing the effectiveness of this approach at improving balance, balance confidence, and balance-related effort in experiments with human subjects.

Acknowledgments

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2.6 Appendix

Gain optimization

We examined the maximum tolerable disturbances in ground height with once per step controllers using two different gain optimization methods, CMA-ES and LQR (Table 2.2). For both gains, ankle push-off work modulation at each step was the most effective method among single individual control method even though the maximum tolerable disturbances differed.

Gain matrices

The gain matrices for the above results are shown in Tables 2.3 and 2.4 for the model with high-gain step length control.

Stability metrics

We also measured two different stability metrics, maximum floquet multiplier and gait sensitivity norm, using the high gain step length control model with controllers designed by LQR. The maximum floquet multipliers were obtained by measuring maximum eigenvalues of the stabilized linear model using each controller at a Poincaré section. The gait sensitivity norms were acquired while the model walked 20 steps after an initial 0.001m step down disturbance. Patterns of maximum

Table 2.2: Maximum tolerable disturbance vs. gain method

Contol Method	Measure (% leg length) CMA-ES	LQR
Step-width	1.10	0.96
Foot placement	1.50	1.50
Ankle roll resistance	0.78	0.12
Ankle push-off work	7.80	2.94
All inputs	7.98	3.21

Table 2.3: Gain matrixes using CMA-ES

Control method	input	Unit	q_{tp}	q_{ap}	q_{ai}	q_{hf}	$\dot{q}_{tp}$	$\dot{q}_{ap}$	$\dot{q}_{ai}$	$\dot{q}_{hf}$
Step-width	qsplay	$\times 1$	0.00	0.00	-0.40	-1.00	0.00	0.25	-10.00	- 0.04
Foot placement	qsplay	$\times 1$	0.00	0.00	-2.81	- 0.08	0.00	0.02	-1.14	-0.01
	phi0	$\times 1$	0.00	0.00	0.01	0.00	0.00	0.00	0.00	-0.01
Ankle roll resistance	Kp	$\times 10^4$	0.15	-0.69	-1.73	1.10	0.26	0.30	-0.53	-0.11
	Kd	$\times 10^4$	-1.86	-1.77	-0.22	-0.06	-0.43	-0.29	-1.15	0.84
Ankle push-off work	Tc	$\times 10^3$	-0.12	-0.51	-1.37	-0.20	-0.09	-0.09	-0.61	-0.04
All inputs	qsplay	$\times 1$	0.00	0.00	-0.32	-0.05	0.00	0.02	-0.03	0.00
	phi0	$\times 1$	0.00	0.00	0.01	0.00	0.00	0.00	0.00	0.00
	Kp	$\times 1$	0.00	-0.20	-0.80	0.50	0.10	0.10	-0.30	-0.10
	Kd	$\times 1$	-0.60	-0.80	-0.10	0.10	-0.30	-0.20	-0.30	0.50
	Tc	$\times 10^3$	-0.17	-0.54	-1.50	-0.20	-0.05	-0.06	-0.75	-0.05

Table 2.4: Gain matrixes using LQR

Control method	input	Unit	q_{tp}	q_{ap}	q_{ai}	q_{hf}	$\dot{q}_{tp}$	$\dot{q}_{ap}$	$\dot{q}_{ai}$	$\dot{q}_{hf}$
Step-width	qsplay	$\times 1$	-0.04	0.69	4.89	0.32	-0.07	-0.03	1.86	0.06
Foot placement	qsplay	$\times 1$	-1.30	-1.95	-0.63	-0.90	-0.56	-0.57	-0.20	-0.14
	phi0	$\times 1$	0.77	1.91	5.29	0.81	0.32	0.36	1.98	0.14
Ankle roll resistance	Kp	$\times 10^4$	-0.23	-0.59	-1.79	-0.25	-0.09	-0.11	-0.67	-0.55
	Kd	$\times 10^4$	-0.06	-0.17	-0.51	-0.07	-0.03	-0.03	-0.19	-0.01
Ankle push-off work	Tc	$\times 10^3$	-0.19	-0.51	-1.52	-0.21	-0.08	-0.09	-0.57	-0.04
All inputs	qsplay	$\times 1$	0.35	0.94	2.89	0.40	0.14	0.17	1.08	0.07
	phi0	$\times 10-5$	0.50	1.31	3.95	0.56	0.21	0.24	1.48	0.10
	Kp	$\times 10^3$	-0.14	-0.36	-1.08	-0.15	-0.06	-0.06	-0.40	-0.03
	Kd	$\times 10^3$	-0.04	-0.10	-0.31	-0.04	-0.02	-0.02	-0.12	-0.01
	Tc	$\times 10^3$	-0.17	-0.46	-1.37	-0.19	-0.07	-0.08	-0.51	-0.03

Table 2.5: Stability measures: maximum floquet multiplier and gait sensitivity norm

Contol Method	Max. floquet multiplier	Gait sens. norm
Step-width	0.513	0.093
Foot placement	0.381	0.073
Ankle roll resistance	0.533	0.114
Ankle push-off work	0.533	0.235
All inputs	0.532	0.308

floquet multiplier and gait sensitivity norm did not match those of maximum tolerable disturbance (Table 2.5).

Table 2.6: Disturbance tolerance with spring-like hip model

Contol Method	Measure (% leg length) CMA-ES	LQR
Step-width	0.13	0.04
Ankle roll resistance	0.28	0.17
Ankle push-off work	0.66	0.41
All inputs	0.81	0.44

Spring-like hip control comparison

We also examined balance restoring abilities for a model with spring-like hip actuation (low-gain proportional control). We examined maximum tolerable disturbances using each high-level controller, designed using both LQR and CMA-ES methods, and found similar trends as in the model with high-gain step length control (Table 2.6); ankle push-off work modulation showed the best performance.

Wider step width model comparison

We examined the balance restoring abilities and energy expenditure for a model with larger step width (0.20 m) and compared results to the model with a normal step width (0.15 m). We designed a once per step ankle push-off work controller using LQR for both models and compared maximum tolerable disturbance and mechanical energy expenditure. The wide step model walked on terrain with random height disturbances of up to 4.3% leg length, 46% higher than with the nominal step width model (2.9% leg length). With ground height disturbances of 2.9% leg length, the wide step width model and nominal step width model spent 130.5 J and 127.9 J of hip energy and 58.3 J and 54.9 J of ankle energy, respectively. The wide step width model was capable to reject a higher disturbance, but spent more energy compared to the nominal step width model.

Slope disturbance comparison

We compared the balance restoring abilities of the foot placement and ankle inversion/eversion control strategies tested here to prior results. To our knowledge, no other studies have applied the same stability metrics. Therefore, we performed additional simulations using disturbances applied in other studies. Using gains derived by LQR, we made the model step down for ten steps using ankle inversion/eversion control or foot placement control only. We found that the model could walk down 2.5° slopes and 0.057 m uneven terrain for foot placement and ankle inversion/eversion control, respectively.

Maximum one-time lateral impulse comparison

We also compared maximum step disturbance in a roll angular velocity to exam the lateral stabilization ability of three step-to-step controllers, which have high lateral direction control authorities; torque offset, step width, and ankle inversion/eversion parameters. We measured the maximum step disturbance by applying incrementally increasing ankle inversion/everion angular velocity disturbance at the first step before the model fell down while walking 20 steps. For the high gain step length control model, the model endured 0.013 rad/s, 0.003 rad/s, and 0.015 rad/s using the ankle push-off work controller, step width controller, and ankle inversion/eversion controller, designed using LQR, respectively. Direct ankle inversion/eversion control showed the highest performance, but the ankle push-off work control also showed a similar stability result.

Chapter 3

Ankle push-off work modulation reduces balance-related effort of individuals with simulated amputation

Inspired by the simulation results, I performed an experiment, studying the effects of the step-to-step ankle push-off work controller on able-bodied subjects with simulated amputation. I found that modulating ankle push-off work on each step reduced metabolic rate and foot placement variability. In addition, subjects tended to prefer the stabilizing controller conditions. The reduction in balance-related effort was not a result of a change in average prosthesis behavior, since average push-off work was maintained across conditions. Instead, changes in energy consumption were due to step-to-step changes in push-off work. The push-off work affects balance during walking in both medio-lateral and fore-aft directions and seemed to help subjects swing their leg more naturally, resulting in reduced metabolic energy consumption. The results suggest that step-to-step ankle push-off work modulation has the potential to reduce balance-related effort in individuals with below knee amputation.

In chapter 4, I performed the study described above with the target population using a modified experimental protocol. This promising result from the able-bodied study further motivated the design of an emulator with an additional degree of freedom (details in Chapter 5) and a balance perturbing device for future studies on stability during human locomotion.

Kim, M., Collins, S. H. (2015) Once-per-step control of ankle-foot prosthesis push-off work reduces effort associated with balance during human walking. Journal of NeuroEngineering and Rehabilitation, 12:43.

This work was presented at:

Kim, M., Tembulkar, T., Collins, S. H. (2014) Modulating prosthetic ankle push-off work at each step reduces balancing effort during walking. Podium presentation at World Congress of Biomechanics.

Abstract

Background *Individuals with below-knee amputation have more difficulty balancing during walking, yet few studies have explored balance enhancement through active prosthesis control. We previously used a dynamical model to show that prosthetic ankle push-off work affects both sagittal and frontal plane dynamics, and that appropriate step-by-step control of push-off work can improve stability. We hypothesized that this approach could be applied to a robotic prosthesis to partially fulfill the active balance requirements of human walking, thereby reducing balance-related activity and associated effort for the person using the device.*

Methods *We conducted experiments on human participants (N = 10) with simulated amputation. Prosthetic ankle push-off work was varied on each step in ways expected to either stabilize, destabilize or have no effect on balance. Average ankle push-off work, known to affect effort, was kept constant across conditions. Stabilizing controllers commanded more push-off work on steps when the mediolateral velocity of the center of mass was lower than usual at the moment of contralateral heel strike. Destabilizing controllers enforced the opposite relationship, while a neutral controller maintained constant push-off work regardless of body state. A random disturbance to landing foot angle and a cognitive distraction task were applied, further challenging participants' balance. We measured metabolic rate, foot placement kinematics, center of pressure kinematics, distraction task performance, and user preference in each condition. We expected the stabilizing controller to reduce active control of balance and balance-related effort for the user, improving user preference.*

Results *The best stabilizing controller lowered metabolic rate by 5.5% (p = 0.003) and 8.5% (p = 0.02), and step width variability by 10.0% (p = 0.009) and 10.7% (p = 0.03) compared to conditions with no control and destabilizing control, respectively. Participants tended to prefer stabilizing controllers. These effects were* *not due to differences in average push-off work, which was unchanged across conditions, or to average gait mechanics, which were also unchanged. Instead, benefits were derived from step-by-step adjustments to prosthesis behavior in response to variations in mediolateral velocity at heel strike.*

Conclusions *Once-per-step control of prosthetic ankle push-off work can reduce both active control of foot placement and balance-related metabolic energy use during walking.*

3.1 Background

People with below-knee amputation experience more falls and lower balance confidence than individuals without amputation (Miller et al., 2001a). Fall risk is more elevated for individuals who report needing to concentrate on each walking step (Miller et al., 2001a), suggesting that difficulty with balance maintenance during steady gait might contribute to increased fall risk. Amputees using passive prostheses expend more metabolic energy during walking (Waters and Mulroy, 1999), which could also be partially due to increases in balance-related effort. For non-amputees, walking on uneven terrain (Voloshina et al., 2013) or with visual perturbations (O'Connor et al., 2012) challenges balance and increases metabolic energy cost. This increase in effort is often due not to changes in average gait mechanics, but rather to changes in step-by-step variations in, e.g., foot placement and associated muscle activity, used for the active control of balance (Bauby and Kuo, 2000). For similar reasons, external stabilization can have an opposite effect (Dean et al., 2007). Among amputees, destabilizing conditions have a much greater detrimental effect on energy cost, walking speed, and perceived effort (Paysant et al., 2006), likely reflecting greater increases in balance-related effort. Such balance-related deficits contribute to reduced mobility, social activity and quality of life for people with amputation (Zidarov et al., 2009). Fall avoidance and recovery training show promise for reducing fall rates among amputees (Esquenazi and DiGiacomo, 2001; Crenshaw et al., 2013a; Kaufman et al., 2014), but are unlikely to reduce the effort associated with active maintenance of balance. Active prosthesis control could complement this approach; in addition to potentially further improving balance confidence and reducing fall rates, enhanced control might also reduce balance-related effort.

Active prostheses have already demonstrated improvements in other aspects of walking performance. Robotic ankle-foot prostheses have been used to reduce metabolic energy consumption during walking by producing more positive mechanical work at the ankle joint than conventional passive devices (Herr and Grabowski, 2012). As the amount of prosthesis work produced during the end of the stance period, or 'push-off', increases, metabolic energy consumption can be reduced (Caputo and Collins, 2014a). Just as average push-off work seems to affect nominal walking effort, perhaps adjustments in push-off work on each step could reduce the effort associated with recovering from small, intermittent disturbances on each step.

Once-per-step push-off work control

Results from recent studies of walking using mathematical models and bipedal robots suggest that once-per-step control of ankle push-off work can improve balance. This approach is based on limit-cycle analysis of gait: at key moments in the gait cycle the system state is sampled, the error from the nominal state (or fixed point) for that instant is calculated, and the error is used to calculate control inputs for the ensuing step. When effective, small changes in control on each step reject small disturbances to the system, improving stability without changing the limit cycle itself. This approach has been used to stabilize two-dimensional walking robots (Hobbelen and Wisse, 2008)

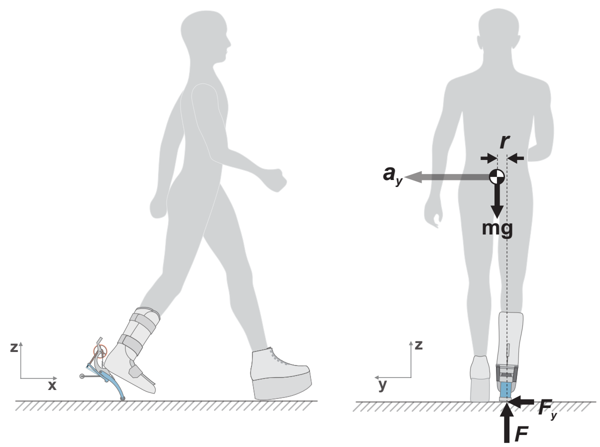


Figure 3.1: Trailing-limb push-off affects both sagittal plane and frontal plane dynamics. Ankle push-off generates a force (F) commonly understood to affect motions in the sagittal plane (*left*) but which also affects motions in the frontal plane (*r*ight). In general, the combination of push-off and gravity, with finite mediolateral displacement between the center of pressure and the center of mass (r) results in a mediolateral force at the foot (F_y), thereby contributing to mediolateral acceleration of the body (a_y). Neglecting rotational inertia about the center of mass, the effect on lateral acceleration is proportional to push-off force as $\Delta a_y = \frac{1}{m} \cdot \frac{r}{L} \cdot \Delta F$, where L is leg length.

including one that set the distance record for legged robots (Bhounsule et al., 2012). We recently used a dynamic model of walking to investigate the effectiveness of once-per-step push-off work control at stabilizing three-dimensional bipedal gait (Kim and Collins, 2013), and found it to be even more effective than foot placement at recovering from random ground height disturbances. This may owe to the fact that push-off affects both frontal-plane and sagittal-plane motions (Figure 3.1). In three-dimensional systems, side-to-side motions tend to be less stable (Kuo, 1999; Donelan et al., 2004; O'Connor and Kuo, 2009), making the effects of push-off on mediolateral velocity especially useful. Another advantage of ankle push-off work control for prosthesis design is that, unlike foot placement strategies, it requires actuation only at the ankle joint. Once-per-step control of ankle push-off work therefore seems like an attractive option for reducing balance-related effort for individuals with transtibial or transfemoral amputation.

Implementing a simulation-based controller in a robotic prosthesis is made challenging, however, by factors such as limited sensory information and model errors. In our simulation study, the best performance was obtained with full state feedback control, in which errors in the position and velocity of all parts of the body were used to make control decisions. This is impractical in hardware. Fortunately, we also found that mediolateral velocity measurements alone could be used to reconstruct desired ankle push-off work within 1% of the value calculated using full state feedback (Kim M, Collins SH: Once-per-step control of ankle push-off work improves balance in a three-dimensional simulation of bipedal walking, submitted). This reduced-order controller retained a substantial portion of the effectiveness of the full-state feedback version, and is more easily implemented in hardware.

A more significant issue is that humans are vastly more complex than the simple models used to derive candidate controllers, which could make the effects of intervention more difficult to observe. Our model included human actuation only at the hips, and treated this as independent from the behavior of the ankle-foot prosthesis (Kim and Collins, 2013). In reality, we expect humans to exhibit complex, neurally-based compensation strategies throughout the body as prosthesis behavior changes, including long term adaptations. The right prosthesis behavior might still be beneficial, of course, if it were to provide a useful component of an overall coordination strategy that involves less effort by the human at steady state. Differences between prosthesis controllers might be difficult to measure, however, since the human could partially compensate for even poor control schemes. To make the effects of push-off work control on balance-related effort more obvious we simulated controllers expected to either stabilize or destabilize the user, and found the expected changes in dynamic stability of the model. A similar relationship might be expected for balance-related outcomes in humans.

Another way of magnifying the effects of prosthesis control on balance-related effort is to make balance more difficult by applying an external disturbance. Human gait exhibits some degree of variability even without explicit disturbances due to internal actuation and sensor noise (Horak et al., 1989; Hausdorff et al., 2001; Collins and Kuo, 2013). When only small external disturbances are applied, the differences in many measures of balance-related effort can be masked by baseline noise. In our simulation model we found that low levels of ground height disturbance caused negligible changes in mechanical work requirements at the hip and ankle (Kim M, Collins SH: Once-per-step control of ankle push-off work improves balance in a three-dimensional simulation of bipedal walking, submitted). A significant external disturbance can make these changes more obvious. A common disturbance encountered by individuals with amputation is ground irregularity (Paysant et al., 2006). This is difficult to implement in a laboratory setting, but a similar effect can be achieved with a robotic prosthesis by applying unexpected changes in the landing angle of the foot at heel strike. This affects the ensuing collision, resulting in significant changes in system energy and both fore-aft and lateral components of center of mass velocity (similar to the effect of push-off illustrated in Figure 3.1). Such a disturbance would therefore be expected to increase active control requirements and balance-related effort.

Measuring balance-related effort

Differences in balance-related effort across prosthesis controllers could be indicated by a combination of step width variability, average step width, within-step center of pressure variability, metabolic rate, cognitive load or user preference. In the present context, 'balance-related effort' refers to the portion of activity associated with balance maintenance during walking, as opposed to activity for 'propulsion', 'body weight support', or other nominal gait requirements. Such effort can be isolated from nominal walking effort if changes are made only in step-by-step prosthesis dynamics, associated with balance, and not to average prosthesis mechanics. Even if the human user were to adjust their average gait mechanics in response to such prosthesis control, for example by taking wider or narrower steps, such changes would primarily relate to changes in balancing strategy and not to the nominal effects of the device.

Step width variability is an indicator of effort arising from active control of foot placement. Subjects tend to increase step width variability in the presence of a disturbance (Collins and Kuo, 2013; O'Connor et al., 2012; Voloshina et al., 2013) and decrease variability with external stabilization (Donelan et al., 2004; IJmker et al., 2013). This suggests increased or decreased use of foot placement control, and associated effort, when balance is challenged or assisted, respectively. If prosthetic ankle push-off control were to make balancing easier for the human user, we might therefore expect to observe reduced active control of step width and reduced variability.

Increased average step width can also indicate an increase in balance-related effort. Humans sometimes increase step width when balance is challenged through sensory-motor impairment (Curtze et al., 2011; Dean et al., 2007) or external disturbances (Voloshina et al., 2013). This strategy, perhaps used to increase 'margin of stability' (Hof et al., 2005), comes at the cost of increased metabolic energy consumption, which increases with the square of step width (Donelan et al., 2001). Our recent simulation study also showed that increasing step width enhanced stability but increased energy cost (Kim M, Collins SH: Once-per-step control of ankle push-off work improves balance in a three-dimensional simulation of bipedal walking, submitted). If prosthesis push-off control were to reduce the need for active balance, this might therefore lead to reduced step width and lower metabolic rate.

Center of pressure variability within the stance phase of each step might also reflect changes in balance-related effort. Strategies based around within-step center of pressure control, including 'zero moment point' control, are widely used to stabilize walking robots (Kajita et al., 2003). In the presence of disturbances to ground height, the center of pressure can be continuously controlled by the ankle joint to maintain balance (Kim et al., 2007). In our recent simulation study (Kim M, Collins SH: Once-per-step control of ankle push-off work improves balance in a three dimensional simulation of bipedal walking, submitted), we found that ankle inversion/eversion torque control could stabilize gait, resulting in a small (about 1%) increase in center of pressure variability. Larger center of pressure variability in the intact limb of individuals with transfemoral amputation suggests that this strategy may be utilized more heavily when other balance pathways are impaired (Hof et al., 2007). With improved prosthesis control, we might expect to find small reductions in center of pressure variability for the intact foot.

Changes in metabolic energy consumption can capture the overall effects of altered muscle activity associated with balance. When people are exposed to significant, random disturbances during gait, their metabolic energy consumption can increase by up to 27% (Paysant et al., 2006; O'Connor et al., 2012; Voloshina et al., 2013). Conversely, providing external stabilization can reduce energy cost by up to 8% (Donelan et al., 2004; IJmker et al., 2013). Such changes are often not associated with altered nominal gait patterns, but rather with step-by-step adjustments in gait mechanics, apparently indicating changes in step-by-step muscular effort associated with balance (Voloshina et al., 2013). If prosthesis push-off control were to supplant a portion of the human user's balance-related effort, we would expect a reduction in metabolic energy consumption.

Walking seems to require the use of some cognitive resources (Ijmker and Lamoth, 2012) and humans appear to divide available resources between walking and other simultaneous tasks (Hollman et al., 2007; Nordin et al., 2010). Individuals with sensory-motor deficits have been observed to sacrifice performance at secondary tasks in an attempt to maintain low gait variability (Hollman et al., 2007), while fall-prone individuals have been observed to pay an energetic penalty (by taking wider steps) so as to maintain both distraction task performance and low gait variability (Nordin et al., 2010). An effective ankle prosthesis controller may therefore result in either improved performance at distraction tasks or greater improvements in other outcomes under distraction-task conditions.

User preference is arguably the most important measure of prosthesis performance, and it strongly correlates with positive reception of a device by consumers (Hafner et al., 2002). Individuals with amputation strongly desire prostheses that positively impact balance (Legro et al., 1999; Hagberg and Brånemark, 2001), and prefer actively-controlled prosthetic knees (Stinus, 2000; Kaufman et al., 2007) that reduce fall likelihood (Highsmith et al., 2010). All other things being equal, we would therefore expect users to prefer prosthesis controllers that contribute to balance maintenance.

Study aims and hypotheses

The goal of this experiment was to examine the effects of once-per-step modulation of prosthetic ankle push-off work on balance-related effort. We hypothesized that appropriate control of ankle push-off work would reduce the effort required to maintain balance during walking, which would be indicated by improvements in some combination of step-width variability, average step width, within-step center of pressure variability, metabolic rate, distraction task performance, and user preference. We hypothesized that an inverse controller would destabilize the user, leading to a deterioration in the same outcome measures. We also tested two baseline conditions, walking in street shoes and walking in the prosthesis simulator without external disturbances, to verify that the use of the prosthesis and the application of external disturbances each increased balance-related effort. We expected the results of this study to inform follow-up experiments among individuals with amputation, eventually leading to the design of prosthetic limbs that reduce balance-related effort during walking.

3.2 Methods

We performed an experiment to investigate how once-per-step control of ankle push-off work affects balance-related effort. We developed a discrete ankle push-off work controller based on a mathematical model (Kim M, Collins SH: Once-per-step control of ankle push-off work improves balance in a three dimensional simulation of bipedal walking, submitted) and implemented it on an existing robotic prosthesis emulator (Caputo and Collins, 2014b) worn by non-amputees using a simulator boot. We conducted a walking experiment with a variety of controllers expected to stabilize, destabilize, or have no effect on the user, while maintaining constant average mechanics. We increased initial balance-related effort by applying a random disturbance to the landing angle of the prosthetic foot and having subjects complete a cognitive distraction task. We also collected two baseline conditions, one with no landing-angle disturbance and the other without the prosthesis. We measured step width variability, average step width, within-step center of pressure variability, metabolic energy consumption, distraction task accuracy, and user preference as indicators of balance-related effort.

3.2.1 Prosthesis control

Hardware platform

We used a tethered, one degree of freedom, ankle-foot prosthesis to implement once-per-step ankle push-off work control. This platform (Figure 3.2, described in detail in (Caputo and Collins, 2014b)) used series elastic actuation and had peak operating torque of 175 N·m, root-mean-squared torque tracking error of 3.7 N·m, peak joint

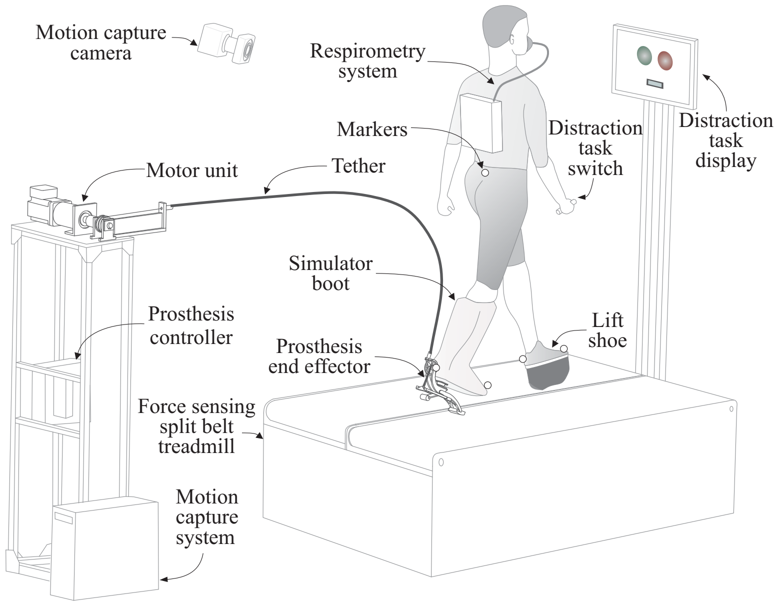


Figure 3.2: Experimental setup. Subjects wore an ankle-foot prosthesis emulator on one leg using an amputation-simulating boot while walking on a force-sensing split-belt treadmill. The prosthesis system was composed of a lightweight prosthesis end-effector, a Bowden cable tether, and a powerful off-board motor and controller. On the opposite limb, subjects wore a lift shoe with a rocker bottom. Reflective markers were attached to the sacrum and the toe and heel of each foot. Marker data was both streamed to a real-time controller and logged by a motion capture system. Subjects wore a wireless respirometry system to measure metabolic rate. Subjects completed a distraction task in which they observed patterns of colors on a monitor and provided responses using a hand-held switch.

power of 1.0 kW, closed-loop torque bandwidth of 17 Hz and prosthesis end-effector mass of 0.96 kg. The system was actuated by a large offboard servomotor and controlled by a high-bandwidth real-time computer (ACE1103, dSPACE Inc., Wixom, MI). Prosthetic ankle angle and torque were measured using onboard sensors.

Mediolateral velocity of the body was measured online using a marker-based motion capture system. A 7-camera system (Vicon, Oxford, UK) measured the positions of a reflective marker attached at the sacrum (Figure 3.2), sampled at a rate of 100 Hz. Lateral velocity of the sacral marker, calculated as the time derivative of sacral marker position, was used to approximate lateral velocity of the center of mass.

Foot contact was determined online using a split-belt treadmill with six-axis force sensing (Bertec Co., Columbus, OH, USA). The sampling rate was 1000 Hz, and data were low-pass filtered at 100 Hz to reduce noise. Foot contact was detected when the vertical component of force was above a threshold value of 20 N. This removed unreliable center of pressure measurements during periods of low force, such as during initial heel contact and just prior to toe off, which could cause artificially high variations in the center of pressure.

Controller design

We implemented once-per-step control of ankle push-off work using mediolateral velocity as a reference. The controller was composed of a high-level discrete controller and a low-level continuous controller.

The high-level controller made adjustments once per step that were intended to stabilize or destabilize the user's gait (Figure 3.3(a)). We calculated the desired magnitude of ankle push-off work as a linear function of the error between nominal lateral velocity and measured lateral velocity, sampled at the moment that the heel of the intact-side foot touched the ground:

$$W_{des} = W^*_{des} + K \cdot (v_{ref} - v_{ml}) \tag{3.1}$$

where W_{des} is the desired ankle push-off work for this step, W^*_{des} is the nominal desired push-off work (approximately equal to the average work over many steps), K is the high-level control gain (with positive values expected to contribute to balance), v_{ml} is the lateral velocity of the sacral marker on this step, and v_{ref} is the reference lateral velocity calculated as a moving average over ten steps (used to prevent changes in average mechanics from affecting balance-related prosthesis control). During pilot tests, we found that not all subjects preferred the same gains, and so we used two magnitudes that seemed to span the most effective range (0.4 and 0.8).

The low-level controller continuously regulated ankle torque as a function of ankle angle so as to deliver the desired magnitude of push-off work over the course

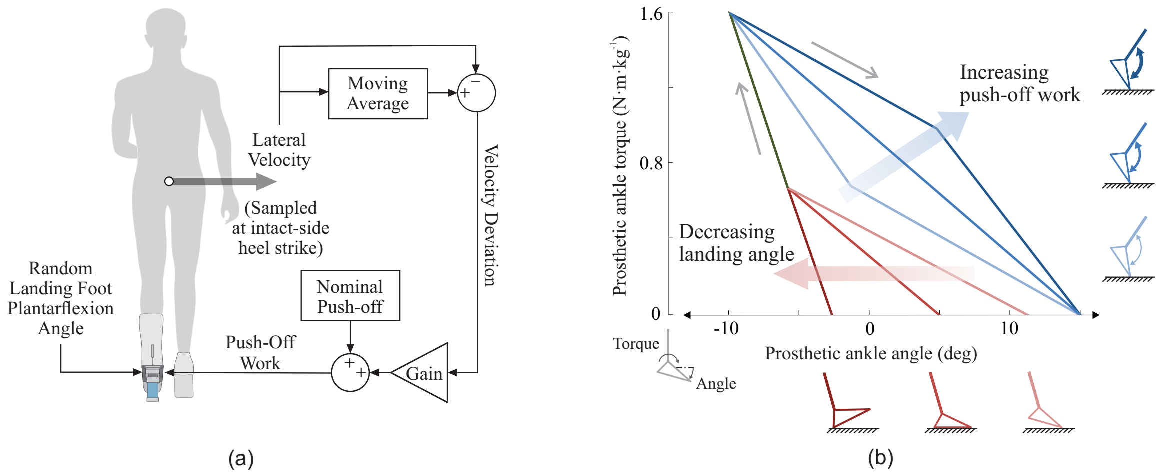


Figure 3.3: Control architecture. (**a**) The high-level controller determined desired ankle push-off work based on mediolateral velocity once per step. Desired push-off work was calculated at the instant of intact-side heel strike, and was equal to a nominal value plus the product of a gain and the difference between lateral velocity on that step and the average lateral velocity over the prior ten steps. Landing-angle disturbances were randomly selected at the beginning of each swing phase. (**b**) The low-level controller continuously regulated ankle torque within each step according to a desired torque-angle relationship. The torque-angle curve was updated by the high-level controller on each step, reflecting changes in desired push-off work (blue portion) and landing-angle disturbance (red portion).

of a step, as described in detail in (Caputo and Collins, 2014a). Desired ankle torque was calculated as a piece-wise linear function of ankle angle, with separate paths for dorsiflexion and plantarflexion phases (Figure 3.3(b)). On each step, the plantarflexion portion of this curve was altered so as to generate the desired magnitude of net push-off work determined by the high-level controller. The plantarflexion torque-angle curve was also adjusted to accommodate differences in peak dorsiflexion angle on each step. The torque control layer then tracked desired torque by rotating an off-board motor (Caputo and Collins, 2014b). During the swing phase, the low-level controller performed position control.

Disturbances

We applied a disturbance in the form of a landing foot angle that was randomly changed on each step. Landing angle was defined as the plantarflexion angle of the prosthesis toe at the moment of foot contact with the ground (Figure 3.3(b)). Landing angle for the next step was randomly selected at the moment the toe lifted off the ground, and the toe was servoed to this configuration during swing. Because of the low inertia of the toe (Caputo and Collins, 2014b) and the cushioning effects of the simulator boot, subjects could not sense differences in toe positioning during swing. Toe angle was maintained until just after the prosthesis toe contacted the ground, as sensed by a spike in ankle torque, at which time the prosthesis switched back into torque control mode. During the ensuing stance phase, the plantarflexion portion of the desired torque-angle curve was adjusted such that the disturbance itself had no effect on net prosthesis work.

3.2.2 Experimental methods

Subjects

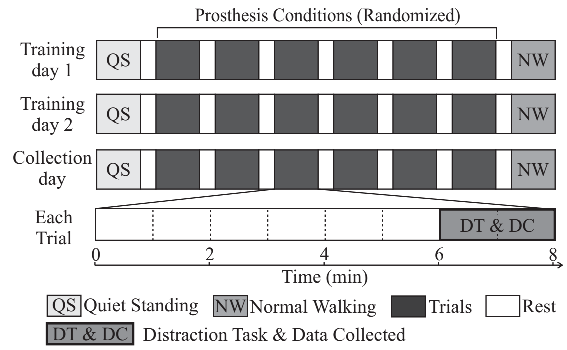
Walking experiments were conducted with able-bodied adults (N = 10 [9 male and 1 female], age = 25 $\pm$4.8 yrs, body mass = 81.2 $\pm$ 5.8 kg, leg length = 0.99 $\pm$ 0.03 m, mean $\pm$ s.d.). Leg length was defined as the distance between markers at the heel and sacrum. To simulate the effects of amputation, subjects wore the prosthesis using a simulator boot and wore a lift shoe on the other leg (Figure 3.2). All participants had prior experience using the prosthesis emulator. All subjects provided written informed consent prior to participating in the study, which was conducted in accordance with a protocol approved by the Carnegie Mellon University Institutional Review Board (HS13-444).

Experimental protocol

Subjects experienced eight conditions per collection (Figure 3.4(a)). Five conditions compared once-per-step push-off work controllers with gains of 0.8, 0.4, 0, -0.4 and -0.8, labeled Stabilizing High Gain, Stabilizing Low Gain, Zero Gain, Destabilizing

Condition Name	Gain	Disturbance	Device
Stabilizing High Gain	0.8	Random Landing Foot Angle	Prosthesis
Stabilizing Low Gain	0.4	Random Landing Foot Anlge	Prosthesis
Zero Gain	0	Random Landing Foot Angle	Prosthesis
Destabilizing Low Gain	-0.4	Random Landing Foot Angle	Prosthesis
Destabilizing High Gain	-0.8	Random Landing Foot Angle	Prosthesis
No Disturbance	0	-	Prosthesis
Normal Walking	-	-	Street Shoe
Quiet Standing	-	-	Prosthesis

(a)



(b)

Figure 3.4: Experimental protocol. (**a**) Each day of the experiment included eight conditions, five of which compared high-level control gains and three of which provided baseline data. During all controller conditions, a disturbance was applied in the form of randomly-changing landing foot angle. In the No Disturbance baseline condition, the high-level gain was set to zero and the disturbance was not applied. In the Normal Walking baseline condition, subjects walked in street shoes without the prosthesis. In the Quiet Standing baseline condition, subjects stood still while wearing the prosthesis. (**b**) Each subject participated in two training days followed by a collection day. Each day, subjects were presented with Quiet Standing, followed by the six prosthesis conditions in random order, and finally the Normal Walking condition. Subjects walked for eight minutes in each trial, followed by three minutes of rest. During minutes six through eight, subjects completed the distraction task. All results presented in the main text are from data collected in minutes six through eight of each trial on the third day.

Low Gain, and Destabilizing High Gain conditions, respectively. The Stabilizing conditions were expected to reduce balance-related effort and the Destabilizing conditions were expected to increase balance-related effort compared to the Zero Gain condition. Landing-angle disturbances were applied in all five of these conditions. Two additional walking conditions provided baseline data. Data were collected for Normal Walking in street shoes and for a No Disturbance condition in which the prosthesis did not apply the landing-angle disturbance. These baseline conditions allowed evaluation of the effects of wearing the prosthesis and applying the disturbance on balance-related effort. Finally, a Quiet Standing condition in which subjects stood still while wearing the prosthesis allowed measurement of resting metabolic rate.

Subjects walked for eight minutes in each walking trial, with three minutes of rest between each (Figure 3.4(b)). A distraction task was performed during the sixth through eighth minutes of each walking trial. Subjects performed all trials in random order, except for Quiet Standing, which was always performed first, and Normal Walking, which was always performed last. Subjects experienced all eight conditions three times on separate days, the first two of which were used for training. All data presented here are from the collection on the third day.

Measures of balance-related effort

We measured metabolic energy consumption, step width variability, average step width, within-step center of pressure variability, distraction task error rate, and user preference. Data were collected during the final two minutes of each trial.

Metabolic energy consumption was obtained through indirect calorimetry using a wireless breath-by-breath respirometry system (Oxycon Mobile, CareFusion, San Diego, CA, USA). Subjects fasted for at least four hours prior to each collection. The rate of oxygen consumption and carbon dioxide production were recorded, and the last two minutes of data were averaged. Steady state oxygen consumption was confirmed by visual inspection. Metabolic rate was calculated using a standard equation (Brockway, 1987) and normalized to body mass. The value for Quiet Standing was subtracted to obtain net metabolic rate.

Step width variability and average step width were calculated using both foot markers and center of pressure data. Step width was defined as the mediolateral displacement between consecutive foot positions. Foot locations were determined at mid-stance, defined as the moment when the sacral marker was directly above the heel marker in the sagittal plane. Marker data and center of pressure data were first low-pass filtered with a cutoff frequency of 20 Hz. We then used the average of the locations of the toe and heel markers at mid-stance to determine marker-based foot position (Collins and Kuo, 2013) and center of pressure location at mid-stance to determine center-of-pressure-based foot position (Donelan et al., 2004). Average step width and step width variability were calculated as the mean and standard deviation, respectively, of all step widths in the corresponding two-minute period.

Within-step center of pressure variability was calculated as the standard deviation of the mediolateral location of the center of pressure at each instant in the stance period. The average center of pressure was subtracted for each step, and center of pressure trajectories were normalized in time to percent stance. At each instant of stance, the standard deviation of center of pressure location across steps was calculated. These values were then averaged across all instants in stance. Center of pressure measurements during initial foot contact or just before toe off are unreliable, but were not included because stance was defined as the period for which the vertical component of the ground reaction force was above a threshold.

Cognitive load was probed by measuring accuracy at a vision-based distraction task for two minutes at the end of each trial. A pair of circles having either the same color (both red or both green) or different colors (red and green or *vice versa*) were shown on a screen (Figure 3.2) every two seconds. Subjects were instructed to press a hand-held button when two consecutive pairs of circles had the same pattern, *i.e.* same followed by same or different followed by different. Error rate was calculated as the percentage of incorrect responses. All subjects reported an ability to distinguish between circle colors. One subject had error rates more than three standard deviations outside the mean, likely resulting from a misunderstanding of the instructions, and their task performance data were removed from the study.

User preference was obtained by asking subjects to rate each condition on a numerical scale. Normal Walking was used as the reference at zero, with -10 corresponding to “unable to walk” and +10 corresponding to “walking is effortless”. Ratings were performed immediately following each walking trial.

A video showing a typical experimental session, including prosthesis hardware and the distraction task, is provided as Additional file 1.

Statistical analysis

We first investigated whether different control gains had any effect on each outcome using repeated measures ANOVA with significance level $\alpha = 0.05$. In cases where significant effects were found, we compared each of the five controller conditions using paired t-tests. We also performed paired t-tests comparing Normal Walking and No Disturbance conditions, to test for an effect of wearing the prosthesis, and between the No Disturbance and Zero Gain conditions, to test for an effect of the disturbance.

3.3 Results

Stabilizing and destabilizing controllers modulated ankle push-off work on each step while maintaining consistent average push-off work. Metabolic energy consumption and step width variability were lower in Stabilizing conditions compared to Zero Gain or Destabilizing conditions. Control gain did not have a statistically significant effect on other balance-related outcomes, but users appeared to prefer Stabilizing conditions. Wearing the prosthesis increased metabolic rate and decreased user preference compared to Normal Walking. The landing-angle disturbance further increased metabolic rate and decreased preference, and also appeared to increase step width variability.

3.3.1 Prosthesis mechanics

The prosthesis applied landing-angle disturbances and modulated ankle push-off work as desired on each step. Landing angles ranged from $-3°$ to $12°$ of plantarflextion across steps (Figure 3.5(a), solid lines). Net push-off work ranged from 0.00 to

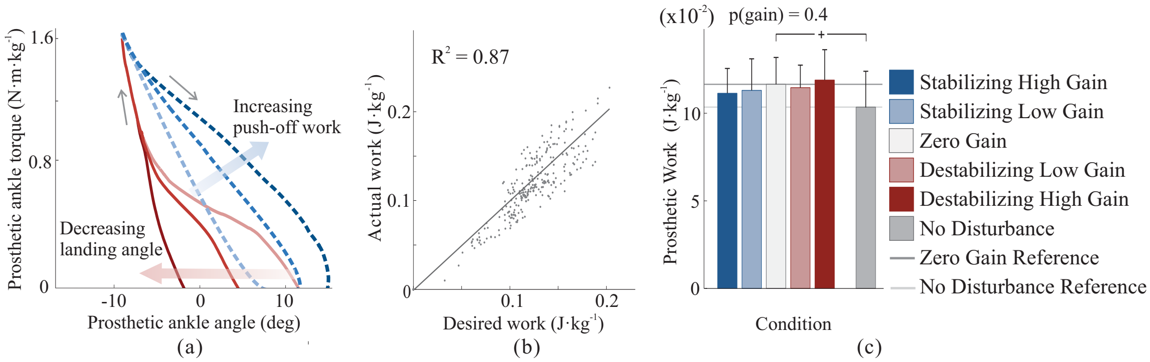


Figure 3.5: Ankle-foot prosthesis mechanics. (**a**) Measured torque-angle relationships for three landing angles and three push-off work values. The red solid lines show the average of all steps in which landing angle was less than 1° (dark line), between 5° and 7° (medium line), and greater than 9° (light line). The blue dashed lines show the average of all steps in which net ankle push-off work was less than 1.3 times the value in Normal Walking (light line), between 1.8 and 2.3 times normal (medium line), and at least 2.8 times normal (dark line). (**b**) The low-level controller closely tracked the desired angle-torque curve, resulting in a strong correlation between desired and measured ankle push-off work on each step. Data are shown for a representative trial. (**c**) Average push-off work remained within 5% of the value for the Zero Gain condition across all other control gains. Subjects received slightly less energy per step in the No Disturbance baseline condition. Blue bars correspond to Stabilizing Gain conditions, white bars to the Zero Gain condition, and red bars to Destabilizing Gain conditions. Darker blue and red bars correspond to High Gains. Light gray bars correspond to the No Disturbance condition. The p-value at top is for a repeated measures ANOVA test for an effect of control gain. Pluses (+) indicate statistical significance among baseline conditions.

0.34 J·kg^{-1} across individual steps, as commanded by the controller (Figure 3.5(a), dashed lines). Desired ankle torque was tracked with root-mean-squared error of 7% across all subjects and conditions, resulting in strong correlation between desired and measured net ankle push-off work across individual steps ($R^2 = 0.87$, Figure 3.5(b)).

Average push-off work did not change significantly across controller conditions ($p = 0.4$). Average net prosthesis work remained within 5% of the value in the Zero Gain condition for all other controller conditions (Figure 3.5(c)). Average prosthesis push-off work appeared to be slightly lower in the Stabilizing control conditions than in the Zero Gain condition.

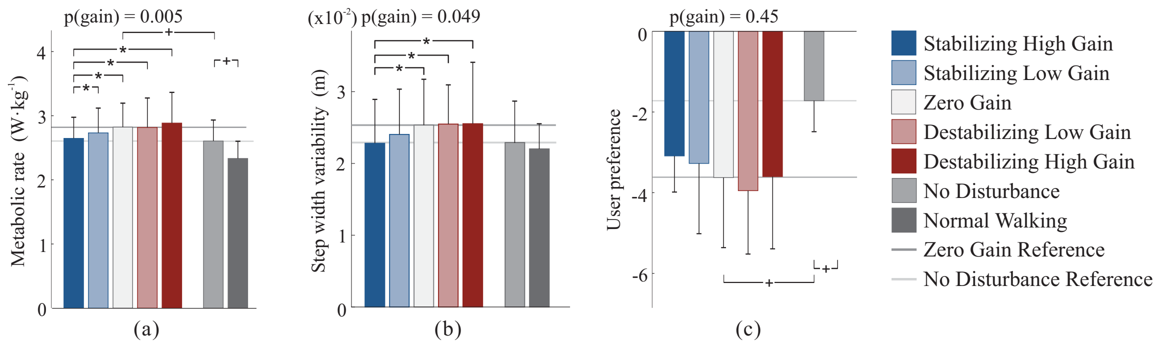


Figure 3.6: Balance-related outcomes. (**a**) Metabolic rate was reduced with Stabilizing control compared to Zero Gain and Destabilizing control conditions. For example, metabolic rate was 8.5% lower in the Stabilizing High Gain control condition than in the Destabilizing High Gain control condition ($p = 0.02$). Wearing the prosthesis increased metabolic rate, as did application of the disturbance. (**b**) Step width variability was lower with Stabilizing control than in Zero Gain or Destabilizing Gain conditions. Wearing the prosthesis appeared to increase step width variability, as did application of the disturbance. (**c**) Subjects appeared to prefer Stabilizing control conditions, although this trend was not statistically significant. Subjects preferred Normal Walking over wearing the prosthesis, and preferred not to have the random landing-angle disturbance. Blue bars correspond to Stabilizing control conditions, white bars to the Zero Gain condition, and red bars to Destabilizing conditions. Darker blue and red bars correspond to High Gains. Light gray bars correspond to the No Disturbance condition, and dark gray bars correspond to the Normal Walking condition. The p-values at top are for repeated measures ANOVA tests for an effect of control gain. Asterisks (*) indicate statistical significance among control gain conditions, and pluses (+) indicate statistical significance among baseline conditions.

3.3.2 Metabolic rate

Control gain significantly affected metabolic rate (ANOVA, $p = 0.005$), with Stabilizing controllers leading to decreased metabolic energy consumption. The Stabilizing High Gain controller reduced metabolic energy consumption compared to all other gains ($p \leq 0.04$; Figure 3.6(a)), including a 5.5% reduction compared to the Zero Gain condition ($p = 0.003$) and an 8.5% reduction compared to the Destabilizing High Gain condition ($p = 0.02$).

Random landing-angle disturbances increased metabolic rate by 9.0%, compared to the No Disturbance condition ($p = 0.02$). Normal Walking required 10.4% less metabolic energy than the No Disturbance condition ($p = 0.0008$).

3.3.3 Step width variability

Variability in step width as measured by center of pressure was affected by control gain (ANOVA, $p = 0.049$), with Stabilizing controllers leading to reduced variability. Stabilizing High Gain control reduced step-width variability by 10.0%, 10.5%, and 10.7% compared to Zero Gain, Destabilizing Low Gain, and Destabilizing High Gain conditions, respectively ($p = 0.009$, 0.046, and 0.030; Figure 3.6(b)). A similar result was observed for step width variability as measured using marker information (Additional file 2: Figure A1).

The random landing-angle disturbance (Zero Gain condition) appeared to increase step width variability by about 10% compared to the No Disturbance condition, but this trend was not statistically significant ($p = 0.2$; Figure 3.6(b)). Walking with the prosthesis in the No Disturbance condition did not increase step width variability compared to Normal Walking ($p = 0.6$).

3.3.4 User preference

Users appeared to prefer Stabilizing control conditions over Zero Gain and Destabilizing control conditions, but this trend was not statistically significant (ANOVA, $p = 0.5$; Figure 3.6(c)). Applying the random landing-angle disturbance (Zero Gain condition) substantially reduced user preference compared to the No Disturbance condition ($p = 0.001$). Subjects preferred the Normal Walking condition over all other conditions ($p \leq 0.007$).

3.3.5 Other outcomes

Within-step center of pressure variability seemed to be reduced by Stabilizing controllers, but this trend was not statistically significant (ANOVA, $p = 0.3$). Wearing the prosthesis appeared to increase within-step center of pressure variability by 14% compared to Normal Walking, and the landing-angle disturbance appeared to increase within-step center of pressure variability by an additional 10%, but neither of these changes were statistically significant ($p = 0.08$ and $p = 0.1$).

Average step width, average stance period and average stride period were unchanged across controller conditions (less than 1.2% change; ANOVA, $p \geq 0.1$). Wearing the prosthesis increased average step width by 30% compared to Normal Walking ($p = 5{\cdot}10^{-7}$), and the landing-angle disturbance increased average step width by an additional 6% ($p = 0.009$) as measured using foot markers, with similar results using center of pressure (Additional file 2: Figure A1).

The rate at which subjects made errors in response to the distraction task was unchanged across controller conditions (ANOVA, $p = 0.3$).

Complete results, including means, standard deviations, and statistical outcomes for all metrics, can be found in the Additional file 2: Figure A1 and Tables A1–A5.

3.4 Discussion

We investigated the effects of once-per-step control of prosthetic ankle push-off work on balance-related effort among non-amputees walking with a prosthesis simulator. We hypothesized that controllers that appropriately modulated push-off work would reduce balance-related effort, while controllers with the opposite effect would increase effort. We found that stabilizing controllers decreased metabolic energy consumption and step width variability, while destabilizing controllers tended to have the opposite effect. Changes were not due to average push-off work or average gait mechanics, which were unchanged across controller conditions. This provides strong evidence that discrete control of prosthesis push-off work can contribute to balance during walking, reducing the need for other balancing strategies such as foot placement, and thereby reducing overall effort.

The primary link between changes in metabolic rate and underlying mechanics seems to be through variability in foot placement. We previously found that once-per-step control of push-off work was effective at stabilizing lateral motions in a three-dimensional model of gait, reducing the need for active control of foot placement (Kim and Collins, 2013). With stabilizing prosthesis control, subjects may have been able to allow more natural leg swing motions, with less need for postural adjustments at heel strike, explaining the observed reductions in foot placement variability. Reduced activity in hip adductors and abductors, implicated in other studies in which balance was made easier or more difficult (Donelan et al., 2004; O'Connor et al., 2012; Voloshina et al., 2013), might account for the observed reduction in metabolic rate. The muscular origins of altered balance-related effort with these controllers could be explored further by collecting electromyographic data in future studies.

Changes in average prosthesis behavior could also affect metabolic rate, but do not seem to be responsible for the changes observed in this study. Average ankle push-off work can have a substantial effect on metabolic rate (Caputo and Collins, 2014a). To avoid confounding balance-related outcomes, we designed the prosthesis controller to have consistent average push-off work regardless of once-per-step control gain. Average push-off work was thereby held within 5% of the value in the Zero Gain condition for all Stabilizing or Destabilizing control conditions. This is a small difference compared to the step-by-step variations in push-off work, which deviated from the average by more than 100% on some steps (Figure 3.5(b)). Stabilizing High Gain control resulted in the lowest metabolic rate but also the lowest average push-off work. Based on a previously established empirical relationship (Caputo and Collins, 2014a), we would have expected this small change in average work to result in a 1% increase in metabolic rate rather than the 5.5% decrease we observed. It is therefore possible that more consistent average push-off work would have further enhanced the benefits of stabilizing control. Subjects also did not change their average step length or step width across controllers, which could otherwise have affected metabolic rate (Zarrugh et al., 1974; Donelan et al., 2001). The observed reductions in metabolic rate, as with step width variability, are therefore best explained by differences in the way push-off work was varied on a step-by-step basis and the effects of such control on balance-related effort for the human.

Changes within baseline conditions also provide insights into the relationships between the use of a prosthesis, external disturbances and balance-related effort. Compared to Normal Walking, simply wearing the prosthesis had a detrimental effect on metabolic rate, average step width, within-step center of pressure variability, and user preference. Some portion of these changes may be due to, *e.g.*, the added mass, height and bulk of the prosthesis simulator boot, but some are likely indicative of increases in balance-related effort from prosthesis use. The addition of a disturbance in landing angle further worsened metabolic rate, average step width and user preference. This suggests that the landing-angle disturbance was effective at increasing balance-related effort, and may have implications for the effects of unpredictable terrain on balance-related effort for individuals with amputation. We separately tested the effect of random changes in push-off work, rather than landing angle, on balance-related effort (Additional file 2: Figure A3), and found that it similarly increased metabolic rate and other indicators of active balance. This provides further support for the idea that step-by-step changes in ankle push-off strongly affect balance.

Pair-wise comparisons of changes in metabolic rate and step width variability did not always yield statistical significance, but our confidence in the reported findings is bolstered by the consistency of the observed changes. Subject-averaged metabolic rate was lower in all Stabilizing control conditions than in the Zero Gain condition, which in turn was lower than in all Destabilizing control conditions. Subject-averaged step width variability, as measured either by center of pressure or marker data, was lower in the Stabilizing High Gain control condition than in all Zero Gain and Destabilizing gain conditions. To further test these relationships, we also examined metabolics and step width variability data from the two minutes before the distraction task was applied, and found the same stratification (Additional file 2: Figure A2(a-c)). The one finding inconsistent with our expectations was that Destabilizing High Gain control appeared to result in reduced step width variability compared to Zero Gain conditions in some cases. This was not consistent with changes in metabolic rate, but was echoed by a trend in user preference. It might be that participants adjusted their balancing strategy in the presence of larger disturbances in ways that were not fully captured by the measures used here. Nevertheless, changes in metabolic rate and step width variability consistently favored the hypothesized effects of push-off control on balance-related effort.

We did not observe statistically-significant changes in mean step width, within-step center of pressure variability, error rates at the distraction task, or user preference across control gains. In some cases, such as with user preference and within-step center of pressure variability, there appeared to be trends resembling those observed in metabolic rate and step width variability, but they were not statistically significant. A greater number of subjects would have allowed validation or rejection of these trends (post-hoc power analyses suggest that an additional forty subjects would have been needed). In other cases, such as with average step width, there were no apparent trends. It may be that subjects relied heavily on foot placement and inversion/eversion control in this task, rather than utilizing a greater margin of stability. The lack of a trend in distraction task error rate is most likely due to a poorly-calibrated task; subjects were approximately 97% accurate in all conditions. Future investigations of cognitive load under similar conditions would lend more insight if they involved a more challenging distraction task.

We did not consider trunk and arm motions in this study, which could have provided an additional resource for balance. Evidence for stabilization strategies using the trunk and arms have been observed in human walking (Benallegue et al., 2013; van Schooten et al., 2011), and variabilities of related measures have been suggested as indicators of stability (Hurt et al., 2010; Moe-Nilssen and Helbostad, 2005). Increased balance-related effort in the arms and trunk might explain increases in metabolic rate despite apparent reductions in step width variability observed in the condition with Destabilizing High Gain control.

We did not have a hypothesis as to which stabilizing control gain would result in greater reductions in balance-related effort, but the observed benefits of the high-gain controller might be explained by subject adaptation. In pilot tests, we observed that subjects with more experience tended to prefer higher gains for the stabilizing controller. We chose two gains that seemed to span the range preferred by both novice and trained users so as to demonstrate some benefit even if little learning occurred. It may be that, by the end of the third day of the experiment, subjects had learned how to best use the stabilizing controller and therefore saw more benefit in the higher gain condition. It is possible that an even higher gain on this feedback loop would have provided experienced subjects with greater reductions in balance-related effort.

Applying the ground disturbance through landing angle of the prosthetic foot was effective in this case, but is not ideal. If there were intrinsic coupling between prosthesis actions related to disturbance and those related to recovery, this could have made balance maintenance easier or more difficult among all control gains. Such a possibility is mitigated by the fact that the disturbance was applied early in the stance phase while stabilizing control actions were performed late in stance. More reassuring is that the disturbance was applied randomly, while once-per-step control was deterministic, meaning that any interactions were likely to wash out over the hundreds of steps measured during the trial. Another concern was the possibility that subjects might predict landing angle based on proprioception. Fortunately, subjects reported that they could not anticipate disturbances, which is supported by increases in balance-related effort when the disturbance was applied. Nonetheless, applying a fully external ground disturbance would avoid the possibility of such interactions and predictions.

Further study will be required to test whether these results are applicable to individuals with amputation. The differences between amputees and non-amputees wearing a simulator boot are numerous, including different levels of training with prostheses and the absence or presence of various sensory and motor control pathways. Perhaps for such reasons, we have previously observed opposite responses to intervention between these populations (Zelik et al., 2011; Segal et al., 2012). Less concerning are the effects of the mass, height and alignment of the prosthesis simulator, since such factors were constant across conditions and are unlikely to interact with once-per-step control gains. While the present results are promising, experiments among individuals with amputation are needed before drawing strong conclusions about effects for this population. Still, with better tuning and more sophisticated control strategies, such as regulation of both lateral and fore-aft body states, such experiments might reveal greater reductions in balance-related effort than observed here.

3.5 Conclusions

We have demonstrated a technique for controlling prosthetic ankle push-off work once per step that reduces balance-related effort during walking in the presence of disturbances. The approach reduces metabolic energy consumption, apparently due to reductions in muscular effort associated with mediolateral foot placement. With small changes, similar control strategies could be implemented in commercially available robotic ankle-foot prostheses. Future work should investigate whether this approach provides similar improvements in balance-related effort for individuals with amputation.

Competing interests

The authors declare that they have no competing interests.

Authors' contributions

M.K. and S.H.C. designed the experiment, M.K. conducted the experiment and analyzed the data, S.H.C. and M.K. drafted and edited the manuscript, and S.H.C. supervised the project. Both authors read and approved the final manuscript.

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Additional file 1: Video of a representative data collection. In this video, a subject walks on the instrumented treadmill with the Stabilizing High controller while performing the distraction task.

Additional file 2: Complete data set and supplementary data. Section 1 and Figure A1 graphically presents all data from the primary study not shown in figures in the main text. Section 2 describes a secondary analysis performed on data from minutes four to six, prior to application of the distraction task, and Figure A2 graphically presents the results from this secondary study. Section 3 describes an additional baseline condition in which push-off work was changed randomly on each step, and Figure A3 graphically presents the results from this additional baseline condition. Section 4 and Figure A4 provide prosthesis mechanics results for the additional analyses and baseline conditions. Table A1 provides mean values for all outcomes in all conditions, and Table A2 provides standard deviations for all outcomes in all conditions. Table A3 provides the results of ANOVA tests for an effect of control gain on each outcome. Table A4 provides the results of paired t-tests comparing control gain conditions, for significantly-affected outcomes. Table A5 provides the results of paired t-tests comparing baseline conditions.

3.6 Additional File 2

3.6.1 Additional measures of balance-related effort

Here we graphically present the measures of balance-related effort that were not included as figures in the main text (Fig. 3.7). In particular, step width variability measured using marker data (rather than center of pressure as shown in the main text) was affected by control gain (ANOVA, $p = 0.03$), with stabilizing control resulting in reduced variability. Step width variability was 10% lower in the Stabilizing High Gain condition than in the Zero Gain condition ($p = 0.03$) and 12% lower than in the Destabilizing Low Gain condition (p = 0.02).

3.6.2 Balance-related effort measured before application of the distraction task

We also analyzed balance-related measures from data taken during minutes four to six of each trial (see cf. Fig. 4 for the trial structure), prior to application of the distraction task (Fig. 3.8). We found similar results to those with the distraction

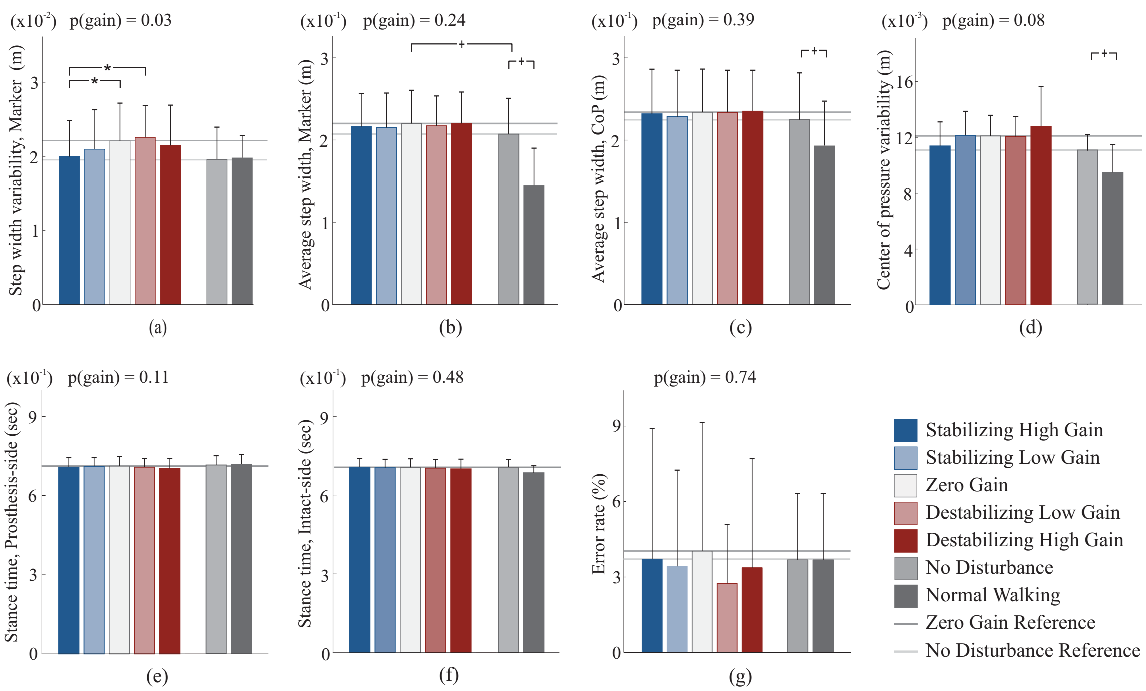

Figure 3.7: Additional measures of balance-related effort. (**a**) Step width variability based on foot markers tracked by a camera-based motion capture system decreased with Stabilizing control. (**b**) Average step width based on foot markers was greater with the prosthesis and with the disturbance. (**c**) Average step width based on Center of Pressure (CoP) measured using an instrumented treadmill was similarly affected. (**d**) Center of Pressure variability within steps seemed to be reduced with Stabilizing control. (**e**) Prosthesis-side stance time was unchanged across all conditions. (**f**) Intact-side stance time was unchanged across all conditions. (**g**) Error rate for the distraction task was unchanged across all conditions. Blue bars correspond to Stabilizing control conditions, white bars to the Zero Gain condition, and red bars to Destabilizing conditions. Darker blue and red bars correspond to High Gains. Light gray bars correspond to the No Disturbance condition, and dark gray bars correspond to the Normal Walking condition. Asterisks (*) indicate statistical significance among control gain conditions, and pluses (+) indicate statistical significance among baseline conditions.

task, reported in the main text. Metabolic energy use without the distraction task was affected by control gain (ANOVA, p = 0.001), with Stabilizing conditions leading to lower metabolic rate. For example, metabolic rate in the Stabilizing High Gain condition was 9% lower than in the Destabilizing High Gain condition (p = 0.008). Step width variability measured using center of pressure was affected

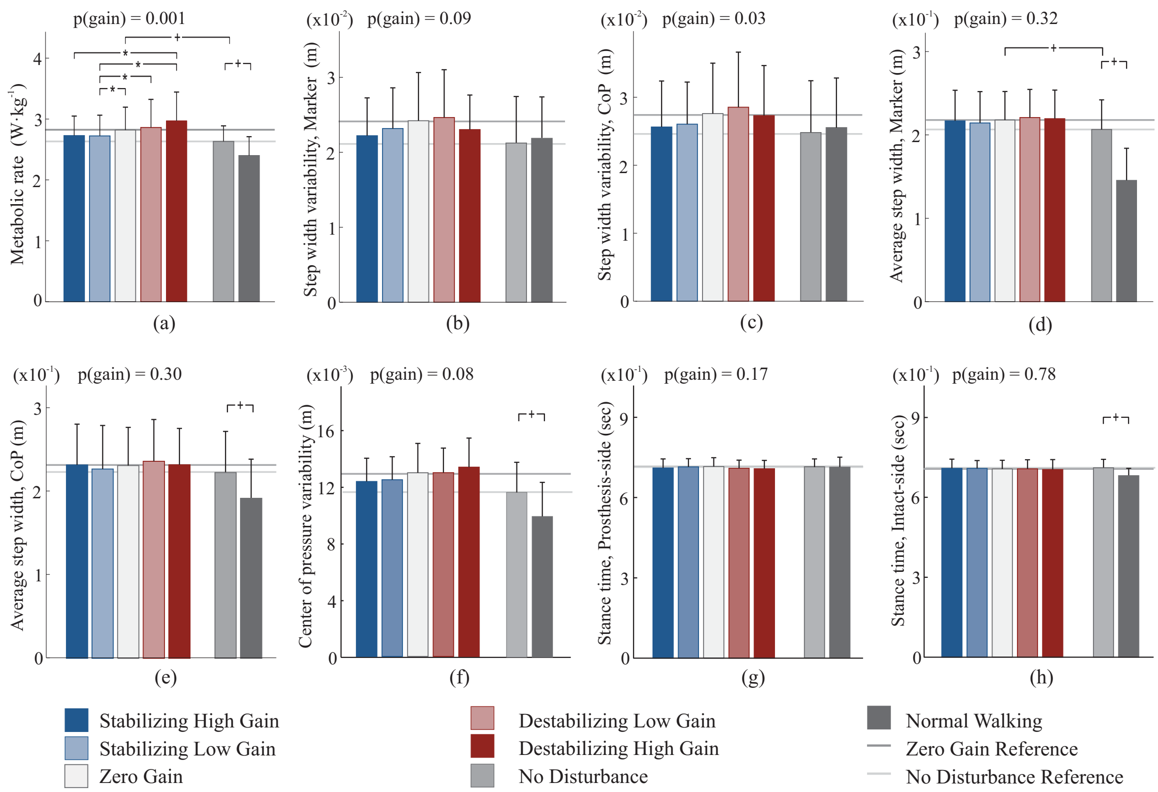

Figure 3.8: Measures of balance-related effort without the distraction task. (**a**) Metabolic rate was reduced by Stabilizing control conditions. (**b**) Step width variability based on foot markers tended to be lower with Stabilizing control. (**c**) Step width variability based on Center of Pressure (CoP) was reduced by Stabilizing gains. (**d**) Average step width based on foot markers was not affected by control gain. (**e**) Average step width based on Center of Pressure (CoP) was not affected by control gain. (**f**) Within-step center of pressure variability tended to be lower with Stabilizing control. (**g**) Prosthesis-side stance time was unchanged across conditions. (**h**) Intact-side stance time was unchanged by control gain. Blue bars correspond to Stabilizing control conditions, white bars to the Zero Gain condition, and red bars to Destabilizing conditions. Darker blue and red bars correspond to High Gains. Light gray bars correspond to the No Disturbance condition, and dark gray bars correspond to the Normal Walking condition. Asterisks (*) indicate statistical significance among control gain conditions, and pluses (+) indicate statistical significance among baseline conditions.

by control gain (ANOVA, p = 0.03), with Stabilizing conditions resulting in lower variability. A similar trend was observed for step width variability measured using foot markers (ANOVA, p = 0.09). Changes in step width variability showed less statistical significance than those during the distraction task period, perhaps because the added cognitive load of the distraction task made prosthesis control more important. Another possibility is that arm motions were affected by holding the clicker used to complete the distraction task, or that the clicker was not held consistently during the first portion of each trial before the distraction task was applied. Baseline comparisons showed similar trends as with the distraction task; wearing the prosthesis (No Disturbance vs. Normal Walking) increased metabolic rate, average step width and within-step center of pressure variability, while the disturbance (Zero Gain vs. No Disturbance) increased metabolic rate and average step width. Other outcomes were not statistically significant.

3.6.3 The effect of randomly changing push-off work on balance-related effort

We tested an additional baseline condition in which push-off work was randomly changed on each step, and measured the same balance-related outcomes both with and without the distraction task (Fig. 3.9). We hypothesized that if push-off work had a strong effect on balance, changing it randomly would strongly increase balance-related effort for the human. We found that random push-off work increased metabolic rate by about 8% compared to the No Disturbance condition (p = 0.02). Random push-off work also increased within-step center of pressure variability (p = 0.04) and reduced user preference (p = 0.007). Other measures of balance-related effort tended to increase with random push-off work.

3.6.4 Average prosthesis push-off work from additional conditions

Average push-off work was unchanged across control gains during the period before the distraction task was applied (Fig. 3.10(a); $p = 0.8$). Application of the disturbance (Zero Gain vs. No Disturbance) slightly increased average push-off work. Average push-off work was not changed by the Random Push-off Work condition, with or without the distraction task ($p \geq 0.4$).

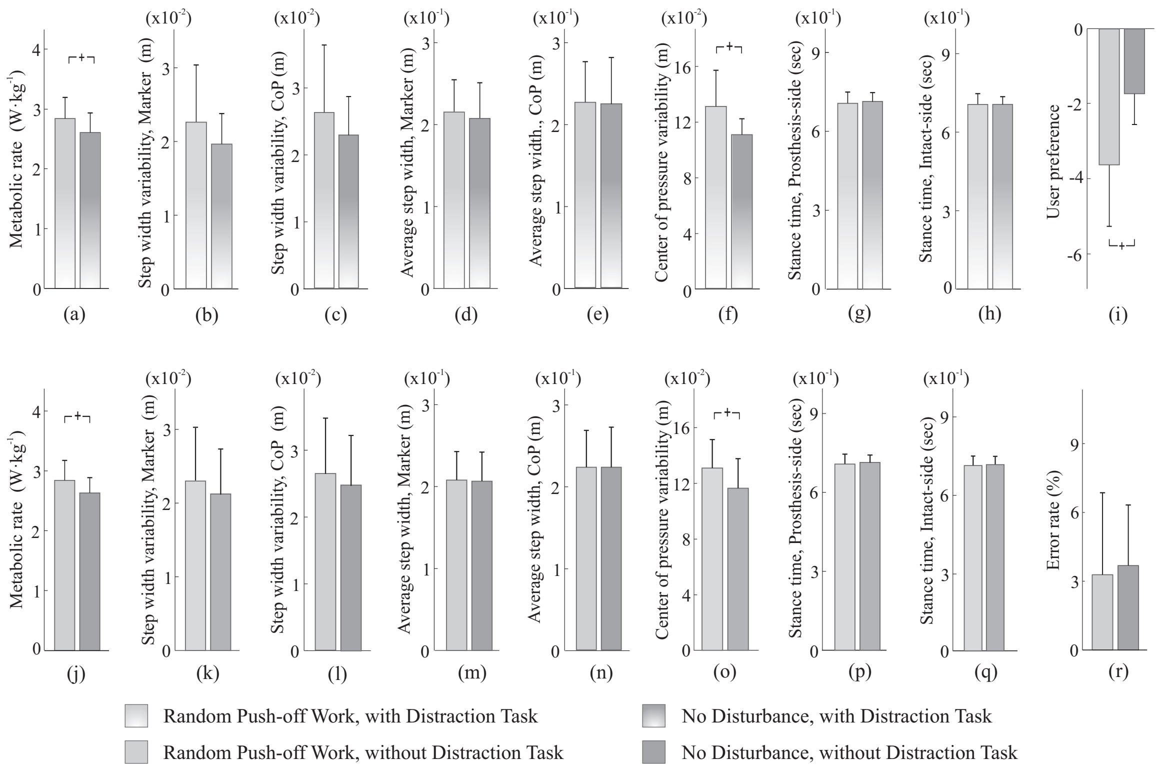

Figure 3.9: The effects of random push-off work on balance-related effort. *Top:* with the distraction task (minutes six to eight). *Bottom:* without the distraction task (minutes four to six). (**a**&**j**) Metabolic rate increased with random push-off work. (**b**&**k**) Step width variability based on foot markers appeared to increase. (**c**&**l**) Step width variability based on Center of Pressure (CoP) appeared to increase. (**d**&**m**) Average step width based on foot markers. (**e**&**n**) Average step width based on Center of Pressure (CoP). (**f**&**o**) Within-step Center of Pressure variability increased with random push-off work. (**g**&**p**) Prosthesis-side stance time. (**h**&**q**) Intact-side stance time. (**i**) User preference decreased with random push-off work. (**r**) Error rate with the distraction task. Light gray bars correspond to the Random Push-off Work condition, and dark gray bars correspond to the No Disturbance condition. Pluses (+) indicate statistical significance (paired t-tests).

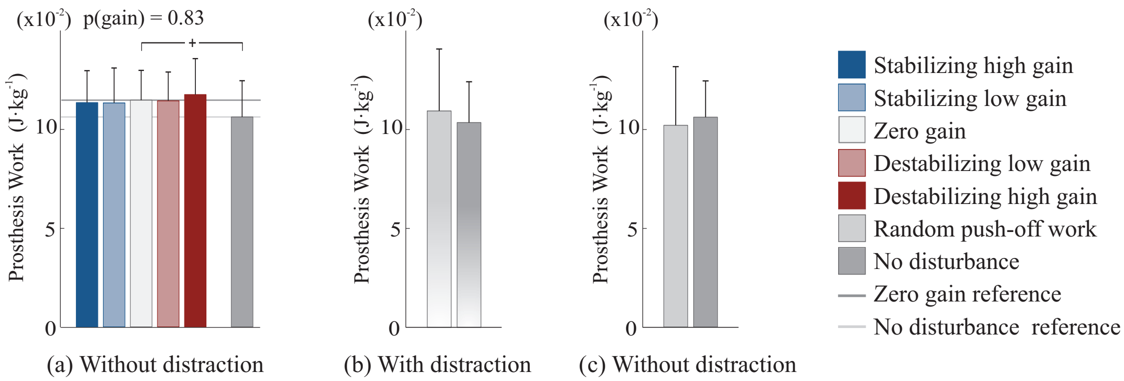

Figure 3.10: Average prosthesis push-off work from additional conditions. (**a**) Push-off work was unchanged across control gains without the distraction task (just as it was unchanged with the distraction task; cf. Fig. 5). (**b&c**) Push-off work was unchanged in the Random Push-off Work condition compared to No Disturbance, with or without the distraction task. Blue bars correspond to Stabilizing control conditions, white bars to the Zero Gain condition, and red bars to Destabilizing conditions. Darker blue and red bars correspond to High Gains. Light gray bars correspond to the Random Push-off Work condition and dark gray bars correspond to the No Disturbance condition. Asterisks (*) indicate statistical significance among control gain conditions, and pluses (+) indicate statistical significance among baseline conditions.

3.6.5 Tables of numerical values

Tables 3.1 and 3.2 provide the means and standard deviations, respectively, of all balance-related outcomes in the study. In these tables, Stab., Destab., Distract., Disturb., var., avg., CoP, and pref. stand for Stabilizing, Destabilizing, Distraction, Disturbance, variability, average, center of pressure and preference, respectively.

3.6.6 Tables of results of statistical analysis of control on balance-related outcomes

The results of repeated measures ANOVA tests for an effect of control gain on balance-related outcomes are presented in Table 3.3. The results of follow-up paired t-tests between controller conditions, only among outcomes that showed a significant relationship, are presented in Table 3.4. Asterisks ($*$) denote statistical significance significance ($\alpha < 0.05$).

Table 3.1: Mean values for all balance-related outcomes

Measurement	Distract. task	Stab. High Gain	Stab. Low Gain	Zero Gain	Destab. Low Gain	Destab. High Gain	Random Push-off	No Disturb.	Normal Walking
Metabolic rate	with	2.647	2.732	2.802	2.820	2.885	2.840	2.605	2.333
($W \cdot Kg^{-1}$)	without	2.727	2.722	2.820	2.860	2.970	2.840	2.632	2.400
Step width var.	with	0.020	0.021	0.022	0.023	0.022	0.023	0.020	0.020
(marker) (m)	without	0.022	0.023	0.024	0.025	0.023	0.023	0.021	0.022
Step width var.	with	0.023	0.024	0.025	0.026	0.026	0.026	0.023	0.022
(CoP) (m)	without	0.026	0.026	0.028	0.029	0.027	0.026	0.025	0.026
Avg. step width	with	0.216	0.215	0.220	0.217	0.220	0.212	0.207	0.144
(marker) (m)	without	0.217	0.214	0.218	0.221	0.220	0.208	0.207	0.145
Avg. step width	with	0.232	0.228	0.234	0.234	0.235	0.227	0.225	0.193
(CoP) (m)	without	0.232	0.227	0.231	0.236	0.232	0.223	0.223	0.192
Within-step	with	0.011	0.012	0.012	0.012	0.013	0.013	0.011	0.010
CoP var. (m)	without	0.012	0.013	0.013	0.013	0.013	0.013	0.012	0.010
Error rate (%)	with	3.656	3.379	3.933	2.099	3.298	3.279	3.682	3.020
User pref.	-	−3.090	−3.275	−3.625	−3.950	−3.600	−3.650	−1.850	0.000

Table 3.2: Standard deviations for all balance-related outcomes

Measurement	Distract. task	Stab. High Gain	Stab. Low Gain	Zero Gain	Destab. Low Gain	Destab. High Gain	Random Push-off	No Disturb.	Normal Walking
Metabolic rate	with	0.328	0.387	0.373	0.459	0.479	0.353	0.329	0.269
($W \cdot Kg^{-1}$)	without	0.321	0.340	0.376	0.462	0.474	0.333	0.255	0.310
Step width var.	with	0.005	0.005	0.005	0.004	0.005	0.008	0.004	0.003
(marker) (m)	without	0.005	0.005	0.006	0.006	0.005	0.007	0.006	0.005
Step width var.	with	0.006	0.006	0.006	0.006	0.008	0.010	0.006	0.004
(CoP) (m)	without	0.007	0.006	0.007	0.008	0.007	0.008	0.008	0.007
Avg. step width	with	0.040	0.042	0.041	0.036	0.038	0.039	0.043	0.045
(marker) (m)	without	0.037	0.038	0.034	0.034	0.034	0.035	0.035	0.038
Avg. step width	with	0.054	0.057	0.053	0.051	0.050	0.049	0.057	0.054
(CoP) (m)	without	0.049	0.052	0.046	0.050	0.044	0.045	0.049	0.046
Within-step CoP	with	0.002	0.002	0.002	0.001	0.003	0.003	0.001	0.002
var. (m)	without	0.002	0.002	0.002	0.002	0.002	0.002	0.002	0.002
Error rate (%)	with	5.192	3.822	5.111	2.349	4.336	3.577	2.641	2.389
User pref.	-	0.896	1.742	1.737	1.571	1.792	1.616	0.755	0.000

3.6.7 Table of results of statistical analysis of baseline conditions

The results of paired t-tests for differences between baseline conditions are presented in Table 3.5. Asterisks ($*$) denote statistical significance significance ($\alpha < 0.05$).

Table 3.3: Results of repeated measures ANOVA tests for an effect of control gain

Measure	Distract. task	ANOVA result
Metabolic rate	with	0.005*
	without	0.001*
Step width var. (marker)	with	0.030*
	without	0.091
Step width var. (CoP)	with	0.049*
	without	0.030*
Avg. step width (marker)	with	0.240
	without	0.320
Avg. step width (CoP)	with	0.390
	without	0.300
Within-step CoP var.	with	0.075
	without	0.074
Error rate	with	0.740
User pref.	-	0.449

Table 3.4: Results of paired t-tests for condition-wise differences among significant outcomes

Conditions Compared		Metabolic rate		Step width var. (marker)		Step width var. (CoP)	
		with distract.	without	with distract.	without	with distract.	without
Zero Gain	Stab. High	0.003*	0.070	0.027*	-	0.009*	0.094
Zero Gain	Stab. Low	0.058	0.018*	0.186	-	0.091	0.234
Zero Gain	Destab. Low	0.802	0.363	0.636	-	0.912	0.586
Zero Gain	Destab. High	0.243	0.053	0.553	-	0.592	0.808
Stab. High	Stab. Low	0.039*	0.911	0.136	-	0.068	0.659
Stab. High	Destab. Low	0.020*	0.063	0.015*	-	0.046*	0.098
Stab. High	Destab. High	0.021*	0.008*	0.055	-	0.030*	0.266
Stab. Low	Destab. Low	0.118	0.049*	0.135	-	0.203	0.082
Stab. Low	Destab. High	0.079	0.011*	0.440	-	0.126	0.272
Destab. Low	Destab. High	0.323	0.141	0.202	-	0.975	0.181

Table 3.5: Results of paired t-tests comparing balance-related outcomes in baseline conditions

Measure	Distraction task	Random Push-off vs. No Disturbance	Zero Gain vs. No Disturbance	Normal Walking vs. No Disturbance
Metabolic rate	with	0.016*	0.011*	0.001*
	without	0.017*	0.028*	0.008*
Step width var. (marker)	with	0.077	0.114	0.875
	without	0.217	0.058	0.579
Step width var. (CoP)	with	0.109	0.156	0.598
	without	0.094	0.130	0.574
Avg. step width (marker)	with	0.330	0.009*	0.000*
	without	0.630	0.009*	0.000*
Avg. step width (CoP)	with	0.764	0.1330	0.001*
	without	0.993	0.184	0.001*
Within-step CoP var.	with	0.046*	0.102	0.084
	without	0.041*	0.184	0.020*
Error rate	with	0.736	0.513	0.621
User pref.	-	0.007*	0.001*	0.000*

Chapter 4

Once-per-step control of push-off work reduces balance-related effort for unilateral, trans-tibial amputees

A follow-up experiment was conducted to evaluate the effect of step-to-step ankle push-off work on balance-related effort of individuals with below knee amputation (N = 4). The controller provided discrete push-off work at each step, as a linear function of the deviation of lateral velocity from a nominal value. In response to this controller, subjects reduced their intact limb control effort during stance to maintain balance. Subject-specific responses to the controller were more fully investigated through the conduction of a single-case experiment involving forced exploration. This experiment was geared towards eliciting insightful details and information about differences in the reactions of each subject to the controller compared to group responses. After forced exploration, some individuals reduced their metabolic rate by at least 10 % compared to the non-stabilizing controller condition. This result suggests that coaching may help people learn more complicated device behaviors, *e.g.* stabilizing side-to-side motion by controlling actuation in the fore-aft direction. Having a better understanding of the behavior of the device after completing the forced exploration session, subjects could more easily reduce balance-related effort.

These results inspired exploring another ankle actuation resource, which directly affected side-to-side motion - ankle inversion/eversion (Chapter 5, 6, and 7).

The contents of this chapter will appear in:

Kim, M., Collins, S. H. (2015) The effect of ankle push-off work control on balance of individuals with below knee amputation, **in preparation**.

Abstract

Individuals with below-knee amputation exert more balance-related effort during walking. Previously, we found that step-to-step modulation of ankle push-off work can reduce balance-related effort for able-bodied subjects with simulated amputation. In this study, we examined the efficacy of the controller with individuals with below-knee amputation ($N = 4$). We hypothesized that this controller would reduce balance-related effort for some individuals. We tested the hypothesis through single-case experimental design, where we presented a neutral controller, then a stabilizing one, followed by the neutral controller again. Amputees developed walking strategies reflective of the complicated balance action of stabilizing controller through forced-exploration periods. To stabilize side-to-side motion as well as fore-aft motion, the stabilizing controller provided push-off work at each step by a linear function of lateral velocity deviation from the nominal value. Single-case experiment results showed three subjects reduced their metabolic rate by 10%, 17%, and 23%, respectively, compared to the neutral one as well as intact limb control effort more than 12%. The statistically significant reduction in intact limb effort ($p = 0.047$) bodes well for the potential of this controller as a balance-restoring method for other amputees. The change in user preference after forced exploration suggests that successful walking with this controller is learned and that it is more positively perceived with training. These results suggest that this step-to-step modulation in push-off work control can be a balance assisting resource with a proper training.

4.1 Introduction

Individuals with below knee amputation experience increased effort to maintain balance during walking (Gates et al., 2013b; Paysant et al., 2006), which may be due to reduced stability. In a previous simulation study, we used a limit cycle walking model to examine the effect of ankle actuation on balance restoration during walking by comparing the ankle actuation's stabilization performance with other stabilization methods (Kim and Collins, 2013). Surprisingly, ankle push-off work modulation stabilized the model during walking as effectively as foot placement control. Inspired by this simulation result, we tested the efficacy of step-to-step ankle push-off work modulation on balance restoration. This evaluation was performed by conducting an experiment with able-bodied subjects wearing an active ankle-foot prosthesis emulator via simulator boot (Kim and Collins, 2015b). The controller reduced balance-related effort, as made evident by reduced metabolic energy consumption and reduced step width variability.

Although these results were compelling, conducting the experiment with able-bodied individuals, rather than the target population, limits interpretation of the study. Individuals with below-knee amputation and their able-bodied counterparts have been shown to respond differently to a given intervention (Zelik et al., 2011; Segal et al., 2012; Quesada et al., 2015). This may be due to inherent differences between the two groups, such as variations in sensory information, motor control ability, and acclimation to walking with a prosthesis. In addition, wearing a simulator boot adds mass to the simulated amputation side and the set up increases the height of both legs, further complicating the comparison. To clarify the uncertainty, it is necessary to test the effect of ankle push-off work modulation on the stability of individuals with below-knee amputation. Experimental design for this purpose, however, is deceptively hard, due to individual's learning ability of a task and the device's indirect behavior to restoring balance.

Learning ability might mask the benefit of the proposed intervention. Individuals uniquely develop a technique to efficiently utilize different devices (Golenia et al., 2014). This ability may enable individuals to find a method to reduce balance-related effort for a neutral controller. Still, we expect that when a controller with destabilizing action is presented, subjects may need to exert more effort to maintain balance if a stabilizing controller can help to reduce the effort. By comparing a stabilizing controller to a destabilizing controller, the potential of a proposed controller may be better revealed.

In addition to the adaptation ability, the fact that this controller indirectly affects side-to-side motion may make learning to use the device more difficult. For such a case, forced exploration can help subjects learn to use the device. This technique has been shown to help able-bodied subjects successfully optimize their gait for a given circumstance (Selinger et al., 2015). Similarly, by providing this exploration period, individuals with below-knee amputation may increase their ability to appropriately use the controller. One possible drawback of this method is the duration of the condition. This technique involves presenting both the forced exploration period and adaptation period in one condition, which requires high physical strength. Individuals with below-knee amputation usually have reduced mobility (Geertzen et al., 2005), which makes it hard to finish such a long condition. Perhaps, by splitting the forced exploration and adaptation period, subjects may be able to complete each condition while learning how to best use the presented controller.

Even with such methods, only some individuals may respond positively to the controller. Each individual has a different learning ability (Golenia et al., 2014; Bouwsema et al., 2010; Vegter et al., 2014), which can be also be affected by a controller (Golenia et al., 2014). Inter-subject variability is even higher for individuals with below-knee amputation (Fey et al., 2010; Wentink et al., 2013). These variability may contribute to different use of average push-off work for each individual with below-knee amputation (Quesada et al., 2015), unlike similar usage of the controller for able-bodied subjects (Caputo and Collins, 2014a). Similarly, even though the group of simulated amputees reduced their balance-related effort by using the step-to-step push-off work controller, only some individuals with below-knee amputation may receive a benefit from the controller.

The positive response of some subjects can be revealed by single-case experimental design. This single-case design focuses more on each individual and enables prompt diagnosis of the effect of the given intervention by examining significance and reliability (Dermer and Hoch, 2012). The evaluation of significant improvement is performed by investigating whether the outcome of an intervention meets a preset objective. The reliability can be visually examined by observing consistent improvement of an intervention, compared to baseline. This comparison can be done using an A-B-A experimental design: initialize with no intervention, introduce intervention and then withdraw intervention with this order. The results of such experiment can show an ordering effect, which can be weakened by averaging baseline conditions with a lag (Gentile and Klein, 1972). The reliability of this controller can be further examined by conducting statistical analysis of randomized order trials of intervention and baseline conditions (Edgington, 1967). However, this randomization prolongs the length of trial. This duration problem can be solved by using random order conditions on different days. The statistical outcome might be more consistent to visual inspection results (Gottman, 1973) by increasing data points using multiple subject data.

While a significant and reliable change in balance-related effort can show external validity of the proposed controller, the applicability to other individuals can be more rigorously evaluated by conducing statistical analysis in a group level. The analysis can be performed using all subject data of single-case experiment if only weak ordering effect is observed (Gentile and Klein, 1972; Hartmann, 1974).

Individuals may receive more benefit from a user-specific stabilizing controller. Each subject learns differently, and fast learners seem to get a higher benefit from a given task (Vegter et al., 2014). The subjects who initially received a benefit may need a higher gain than others to get a maximal assistance from a device. In addition, considering continuous learning ability for an intervention (Selinger et al., 2015), the subject and others may eventually receive more benefit from a slightly more aggressive controller. The gains might be able to be estimated by comparing results between a stabilizing controller and a destabilizing controller from the previous trials.

To gauge the efficacy of the different controllers at improving balance, a collection of balance-related measures can be used including step width variability, metabolic energy consumption, step width average, within a step center of pressure variability, and user preference, similar to our previous study (Kim and Collins, 2015b). We used step width variability to explore the costs associated with foot placement control, step width average to show the effort required to modify nominal behavior, intact limb center of pressure variability within step to reveal the costs associated with ankle inversion/eversion control of the intact limb, and metabolic energy consumption to indicate changes in muscular activity adopted to maintain stable walking. Finally, we measured user preference to show personal preference of each controller condition.

The goal of this study was to investigate the effects of once-per-step modulation of ankle push-off work on balance-related effort in individuals with below-knee amputation. We hypothesized that a controller with step-to-step stabilizing ankle push-off work control would reduce balance-related effort in some individuals with below-knee amputation and a controller with the opposite gain would increase this effort. We also hypothesized that each individual would differently adapt to a controller, and the single-case experimental design with forced exploration would reveal the efficacy of the controller for each individual. Balance-related effort was evaluated from a combination of balance-related measures.

4.2 Methods

We conducted the experiment with four individuals with below-knee amputation. They completed three sessions including a session for single-case experiment with forced exploration. For regular group experiments, subjects experienced two

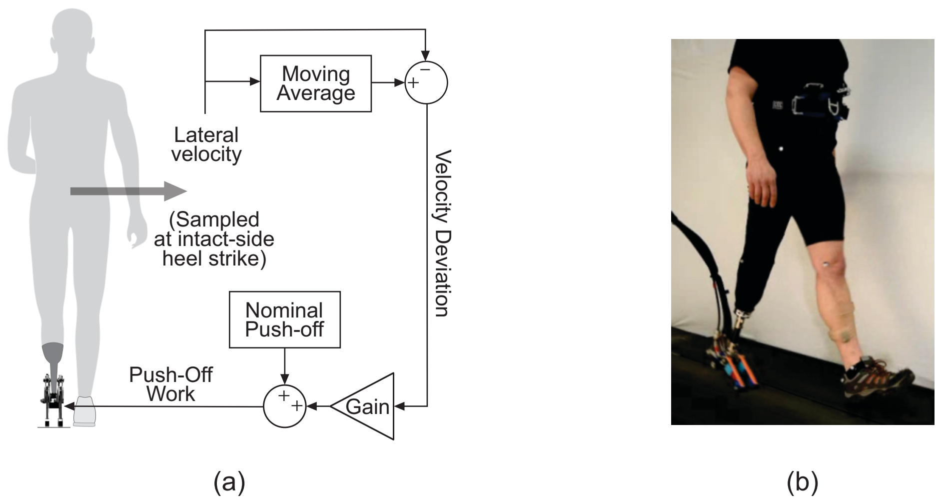

Figure 4.1: (a) Ankle push-off work was determined at each step by multiplying a gain with the measured lateral velocity deviation from nominal values. (b) We conducted experiment with individuals with below knee amputation.

stabilizing controllers, one neutral controller, and two destabilizing controllers. For the single-case experiment, subjects were exposed to the neutral controller and stabilizing controller in a prescribed order. Metabolic rate, step width variability, average step width, center of pressure variability on the intact limb side, and user preference were measured to capture balance-related effort (Detailed description in (Kim and Collins, 2015b)).

4.2.1 Prosthesis control

Experimental hardware

We implemented the ankle push-off work controller on a recently developed two degree-of-freedom prosthesis (Collins et al., 2015a). This device is lightweight (0.72kg), yet can provide high plantarflexion torque (up to 180Nm) with a high bandwidth (20Hz) in a linear range. These characteristics make the emulator ideal for testing control ideas. Additionally, an instrumented treadmill (Bertec Co. Columbus, OH, USA) was used to obtain multi-directional force information.

Controller design

The controller was composed of a high-level controller and low-level controller, similar to our previous study (Kim and Collins, 2015b). The high-level controller is a discrete step-to-step controller, which determined the push-off work at each step by multiplying a gain with the deviation of the current velocity from the nominal velocity at opposite limb heel strike (Kim and Collins, 2015b)(Fig. 4.1)). The velocity was calculated by summing lateral force, dividing the force by mass, and then integrating the acceleration in time. Heel strike was detected when the force was above a defined threshold. The low-level controller continuously actuated ankle plantar flexion and ankle inversion/eversion torque to deliver desired work at each step. The average work was controlled to deliver the same amount work as in the first trial by applying discrete proportional and integration controller on work.

4.2.2 Experimental design

Participants

Experiment was conducted with four subjects with below knee amputation (N = 4, all male, all traumatic, all K3 ambulators, 3 left-side amputation, age = 47.8 ± 14.3 years, body mass = 79. ± 12.7 kg, height = ± 0.037 m, mean ± s.d.). All participants have previously participated in studies with this device (Kim et al., 2015). Experimental protocol approved by the Carnegie Mellon University Institutional Review Board.

Experimental protocol

The experiment was composed of three sessions: an acclimation period, a group experiment and gain selection period, and a single-case experiment with forced exploration period. For the acclimation and group experiment periods, subjects completed five step-to-step push-off work controller conditions. For the single-case experiment session, subjects experienced Stabilizing controller with predetermined gain and Neutral controller.

For all sessions, subjects finished two baseline conditions: a quiet standing condition and a prescribed prosthesis condition. During the quiet standing period, subjects stood naturally while wearing the robotic ankle-foot prosthesis for three minutes. This condition was conducted before subjects walked on the robotic ankle-foot prosthesis. For the prescribed condition, subjects walked on their device. This prescribed condition was randomly positioned at the beginning or end of the experimental protocol.

The speed of walking was set to 1.25 $m \cdot s^{-1}$, but for less active subjects, we adjusted the speed between 0.7 $m \cdot s^{-1}$ and 1.25 $m \cdot s^{-1}$.

Day 1 - Acclimation period: Each subject participated in for seven trials in total, one standing rest condition, one with their prescribed prosthesis, and five with the robotic prosthesis operating under five controller conditions. The five controller conditions are High gain stabilizing, Low gain stabilizing, Neutral, Low gain destabilizing, and High gain destabilizing. The normalized gains for the conditions were $-50, -25, 0, 25, 50$ $(m \cdot s^{-1})^{-1}$, respectively. The Neutral, Stabilizing, and Destabilizing step-to-step controllers were block randomized throughout the sessions. For each stabilizing and destabilizing controller block, we presented controllers in order of increasing absolute gain. Each trial lasted 5 minutes and between trials, there was a 4-minute break. For less active subjects, we adjusted the trial length and rest length between 4 - 5 minutes, and 4 - 10 minutes, respectively.

Day 2 - Group experiment and gain selection period: The protocol was the same as the first period, except the order of the controller conditions was fully randomized.

Day 3 - Single-subject experiment with forced exploration period: For the third day of experimentation, the subjects completed six conditions walking on the controlled prosthesis, one condition on their prescribed prosthesis and participated in a period of standing rest. For the six controller conditions, the Stabilizing controller (B) and Neutral controller (A) were presented with fixed A-A-B-B-B-A order in order to examine the effect of the controller and exploration. First, the Neutral controller was presented for four minutes with forced exploration. Then, this controller was presented on its own for five minutes while collecting data. Then condition B was presented with forced exploration for four minutes. This was followed by two uninstructed periods of condition B, the first lasting four minutes to provide further training and the second lasting five. We collected data for the second uninstructed period. The final condition was with controller A for five minutes while collecting data. For less active amputees, we reduced the data collection time to between four and five minutes. Between trials we provided a rest of four to ten minutes depending on activity level.

During this day of collection, a pre-determined gain calculated from the user's previous trial reaction was used with the stabilizing controller. The response of the subject was compared to both a destabilizing and a stabilizing condition. We used a 10% reduction in metabolic rate and one other balance-related effort measures (user preference, step-width variability, center of pressure variability in intact limb, or average step width) to determine when we had reached the optimal gain. We used this value if the gain was the highest. A 30% higher gain was used for the experiment with forced exploration if the gain was the lowest. A 30% higher gain than the low gain was used in the event that the metabolic rate showed a reduction of more than 10% for the stabilizing controller condition rather than the destabilizing controller condition. We proceeded to analyze the acclimation period data if a trend was not clear after the first collection period. Likewise, if a more than 10% reduction in the stabilizing controller condition metabolic rate was noted when compared to the destabilizing high controller condition, a gain 30% higher than the low gain was applied. The lowest gain was implemented when no reduction in metabolic rate was recorded with a subject.

During forced exploration, subjects received verbal instruction to try different movements every 15 seconds. The instructions were provided for the final 3 minutes of the trial and consisted of the following verbiage: "lean a little bit to your left," "lean a little more to your right," "sway more side to side," "keep your body upright, use limited sway," "lean a little forward," "lean a little backward," "take slightly wider steps," "take slightly narrower steps," "take slightly longer steps," "take slightly shorter steps," "swing your arms more than you typically would," and "swing your arms less than you typically would." For less-active individuals with below knee amputation, instruction was provided less frequently (every 20 seconds) to allow more time to acclimate to the condition. Additionally, "keep your body upright, use limited sway," "take slightly shorter steps," and "swing your arms less than you typically would" were eliminated.

4.2.3 Balance-related measurements

We measured metabolic energy consumption, step width variability, average step width, within-step center of pressure variability, and user preference. We collected data during the final two minutes of each trial. Metabolic energy consumption was calculated via indirect calorimetry. Step width variability and average step width were calculated using five motion capture markers (left heel, right heel, left toe, right toe and sacrum) and treadmill force-plate information. Within-step center of pressure variability in intact limb side was calculated after normalizing data over time during stance phase and calculating standard deviation at each instant of time. At the end of each condition, we asked subjects to rate the condition on an absolute scale. Detailed explanation on experimental setup can be found in (Kim and Collins, 2015b).

4.2.4 Data analysis

Data analysis for each subject

The Stabilizing controller's significant contribution on balance-related effort and reliability were examined by analyzing single-case experiment session. The significance of the contribution was examined by comparing averaged Neutral controller conditions and the Stabilizing controller condition. We considered that the controller significantly lowered effort if the metabolic rate reduction was more than 10% and moderately reduced effort if the difference was more than 6%, based on previous experiment results (Kim and Collins, 2015b). Additionally, for the subject who reduced overall effort, the cause of the lowered overall effort was investigated by examining correlation between metabolic energy consumption and other balance-related measures. We also examined training effect on reduced balance-related effort by conducting three-way ANOVA for the subjects who reduced metabolic rate with significance level of 0.05.

Reliability was visually inspected by observing lowered balance-related effort indicator for the Stabilizing controller condition compared to the Neutral controller conditions. We further inspected reliability for the responded subjects who reduced metabolic energy consumption by conducting statistical analysis across three days between Neutral controller and stabilizing controller conditions. Averaged outcomes of stabilizing controllers were used for the acclimation and group experiment periods. We used two-way ANOVA with significance level of 0.05. We also conducted three-way ANOVA to examine the effect of the training.

External validity

External validity was evaluated by conducting statistical analysis of the Stabilizing controller condition, and averaged Neutral controller conditions to weaken ordering effect caused by fixed order trials on single-case experiment period. We further tested the validity by conducting a two-way ANOVA test with a linear model with a significance level of 0.05 using group experimental period data. Once the statistical significance was found, we further conducted paired t-test by comparing each controller condition to the Destabilizing high gain controller. We also performed a post-hoc power analysis.

4.3 Results

4.3.1 Prosthesis mechanics

The controller modulated ankle push-off work at each step, proportional to the deviation between lateral velocity and nominal velocity while maintaining average torque across trials. For the destabilizing controller condition, the average minimum and maximum value of the net push-off work varied -3J to 13J with an average of 6J. Average net prosthesis work remained within 2% of the work value in the zero gain condition ($p > 0.7$). Desired torque was tracked within a root-mean-square error of 6% across trials.

4.3.2 Balance related outcomes

Data analysis for each subject

Individually, three subjects reduced metabolic rate for the Stabilizing controller by 10%, 18%, and 23% (Fig. 4.2(b)) and lowered intact limb control effort by 13%, 15%, and 12%(Fig. 4.2(a)). Day to day analysis of three subjects showed that the subjects significantly reduced metabolic rate across days ($p = 0.004$ without considering training effect, $p = 0.067$ considering training effect). Their reduction seems to be correlated with each subject's intact limb control effort and foot placement control effort ($p < 0.0016$ and $p < 0.0028$) (Fig. 4.2). One subject increased his metabolic energy consumption by 6% and reduced center of pressure variability by 8%.

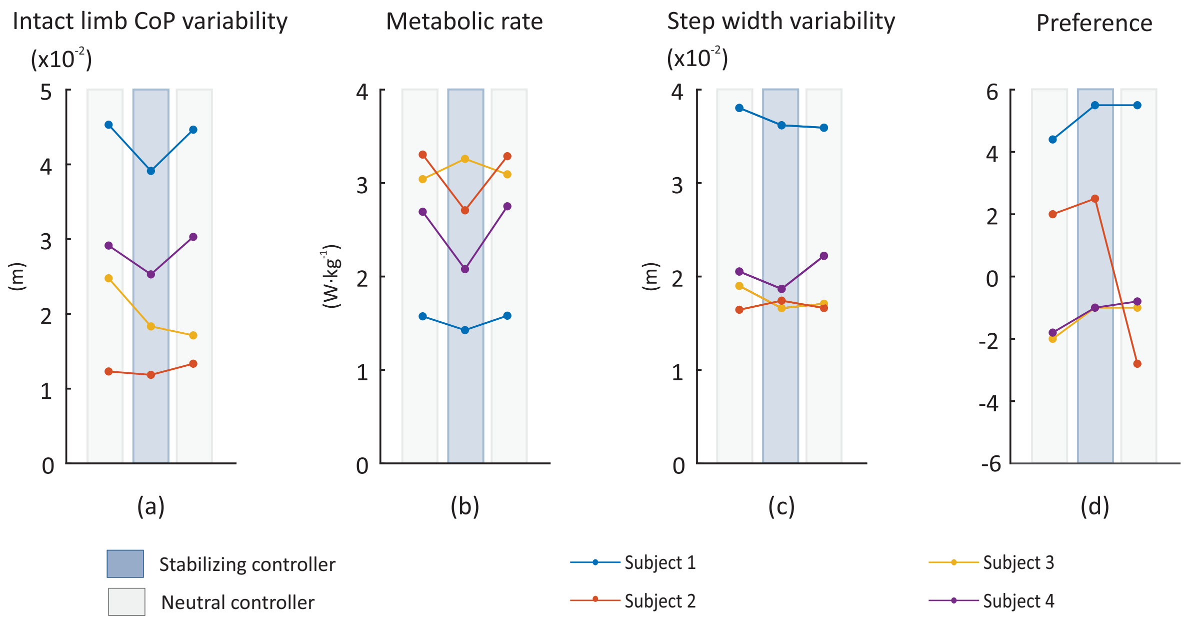

Figure 4.2: Balance related outcomes on single-subject experiment with forced exploration period. Three subjects reduced intact limb control effort and metabolic rate by more than 10%. Subjects also started to show a favorable trend of preference in balance for the Stabilizing controller.

After the forced exploration period, three subjects reduced their metabolic rates from 1% to 10%, from 9% to 18%, and from 28% to 23%, respectively. Across three subjects, this training effect significantly affected reduction in metabolic rate ($p = 0.012$), intact limb control effort ($p = 0.036$), and foot placement control effort ($p = 0.008$). Training also affected each subject's opinion of the different controllers ($p = 0.013$).

External validity

Single-case experiment results show that on average, center of pressure variability on the intact limb side was 13% lower for the Stabilizing controller condition compared to the the Neutral controller condition ($p = 0.047$, power = 77%) for the single-case experiment period (Fig. 4.3(a)). Metabolic rate also presented high difference in level (11%), but it was not statistically significant across subjects ($p = 0.23$) (Fig. 4.3(b)). Post-hoc analysis showed that 13 subjects are necessary to reveal clear

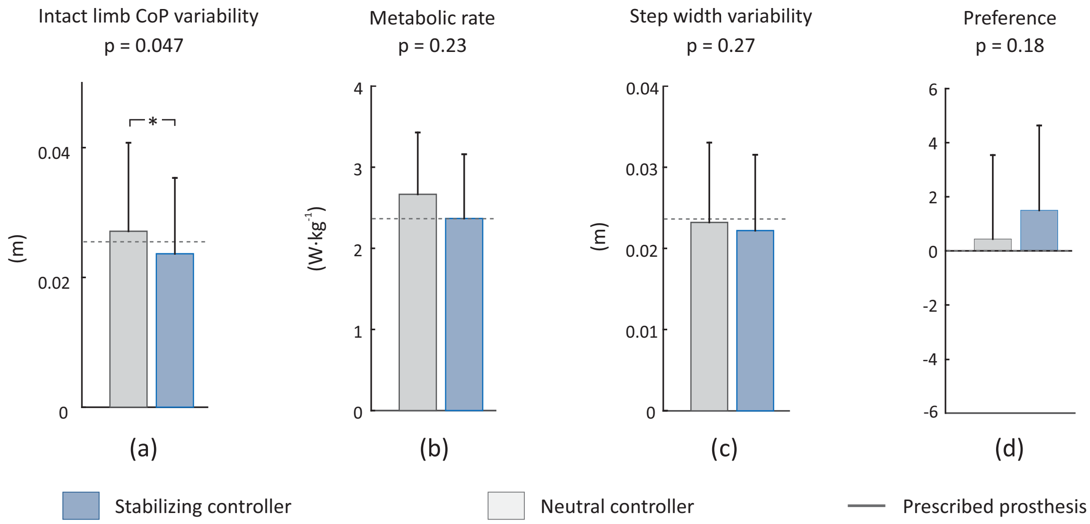

Figure 4.3: Subjects statistically significantly reduced intact limb control effort and showed reduced metabolic energy consumption trend.

results. Subjects consumed almost the same metabolic energy for the Stabilizing controller condition compared to their prescribed prosthesis (difference = 0.13%, p = 0.98) (Fig. 4.3(b)). Step width variability seemed to decrease by 4% for the Stabilizing controller condition (p = 0.27) (Fig. 4.3(c)). Subjects seemed to prefer stabilizing controller conditions (p = 0.18) (Fig. 4.3(d)). Average step width did not show a trend (p = 0.26).

Group experiment results only show some trend in metabolic rate. Metabolic rate seemed to decrease for the stabilizing controller condition (ANOVA, p = 0.1). The reduction was 11% lower for the Stabilizing high condition compared to Destabilizing high condition. To avoid a Type II error, at least 10 subjects were necessary. Average step width appeared to be affected by controller condition (ANOVA, p = 0.2). Average step width was 3.6% lower for the Stabilizing high condition than Destabilizing high condition. Step width variability, Center of pressure variability, user preference did not significantly change across controller conditions (ANOVA p = 0.75, 0.46, and 0.75, respectively).

4.4 Discussion

We hypothesized that some subjects would reduce balance-related effort when the step-to-step ankle push-off work controller was provided. The controller produced a wide range of push-off work while maintaining average push-off work across different conditions. In response to the controller, three subjects reduced their metabolic rate and intact limb effort more than 10%. Only one subject increased energy consumption by 6%. Statistically significantly reduced intact limb control effort by 13% suggests that this controller also can provide a similar benefit to other amputees. The benefit seems to be realized more easily with forced exploration.

The controller reduced the subjects' intact limb control effort and tended to help lower foot placement effort. When testing this controller, in the previous study with able-bodied subjects with simulated amputations, we saw reduced intact limb control effort, but the result was not statistically significant (Kim and Collins, 2015b). The able-bodied subjects reduced step width variability with statistical significance. On the other hand, in this study, individuals with below-knee amputation statistically significantly lowered the intact limb control effort, but their reduction in step width variability was not statistically significant. Since individuals with amputations rely on their intact limb more than their able-bodied counter parts to maintain balance (Hof et al., 2007), this controller might be more effective at reducing intact limb balance-related effort for people with amputations.

During the single-case experiment, three subjects showed a reduction metabolic rate for the stabilizing controller. These results were not random ($p = 0.004$ across days). In addition, they appear to learn how to better use the provided controller during the forced exploration period, shown by about 10% further reduction in metabolic rate. This reduction seemed to be caused by learning to reduce intact limb control effort ($p = 0.04$) and foot placement ($p = 0.008$) across days, regardless of subject. The forced exploration period also affected each subject's opinion on the provided controller ($p < 0.04$).

This reduction was not caused by average push-off work. Change in push-off work can affect overall effort. However, in this study, we more tightly maintained the average push-off work, compared to previous study, by applying discrete proportional and integration controller on push-off work. For each single-case experiment, the push-off work was maintained within 0.1% ($p = 0.8$). This result suggests that the change in average work did not contribute to the reduction of intact limb control effort and high metabolic rate reduction in three subjects. In actuality, the reduction was caused by changes in step-by-step push-off work.

Despite of three subjects' reduced metabolic rate, we did not see a statistically significant reduction in metabolic energy consumption across all subjects. Different responses in metabolic rate may be partially explained by inter-subject variability of learning the proposed controllers. The process of learning a new task or device is different among individuals within a tight age group (21 $\pm$ 1.68 years) (Golenia et al., 2014). The dissimilarity in learning demonstrated in this study (Van Hedel and Dietz, 2004) might be amplified by high inter-subject variability in individuals with amputations (Fey et al., 2010; Wentink et al., 2013) and a wide range of ages (Nagai et al., 2011; Krasovsky et al., 2014), compared to our previous study (Kim and Collins, 2015b).

Another possible explanation for the diverse responses to overall metabolic energy consumption could be high individual variability in muscle activity (Fey et al., 2010; Wentink et al., 2013; Nagai et al., 2011). For example, individuals may optimize their muscle activity in time (Moore and Marteniuk, 1986) in a distinct manner. These muscle activity differences may explain a subject's increase in metabolic rate for the stabilizing controller condition, as well as the wide range of reduction magnitude among subjects. This diverse level of reduction also contributed to an increase in p-value. These variability in learning and muscle activity may have caused the lack of a statistically significant trend in metabolic rate.

For rapid responders, who reduced their metabolic energy consumption, measuring upper body motion may reveal more about their initial balance control strategy. Even though they reduced their metabolic rate in the initial period, they did not reduce lower limb control effort. This might be caused by non-optimized muscle activity (Moore and Marteniuk, 1986), or upper body motion to maintain balance (Beurskens et al., 2014; Curtze et al., 2011). By measuring upper body motion using a motion capture system, their initial balance control strategy may be more clearly revealed.

More training with forced exploration may provide benefits from the step-to-step push-off work controller. Training sessions are essential in learning how to use a controller (Selinger et al., 2015), and the time duration of the training session may vary depending on difficulty of a task. In our experiment, the subjects who did not reduce metabolic energy consumption for the stabilizing condition also lowered their intact limb control effort and foot placement effort after forced exploration. The ankle push-off work modulation is a complicated control action because it indirectly affects side-to-side balance. Perhaps, with a longer forced exploration period and training, such subjects might also be able to reduce their overall energy consumption and improve their opinion on controller.

An alternate assistance strategy is providing a controller that is easier for individuals with below-knee amputations to understand. The provided step-to-step ankle controller greatly reduced metabolic rate for some subjects, but others seemed to get less benefits and may have needed more training. Perhaps, these inconsistent results were caused by the indirect control characteristics of the provided controller - impacting coronal plane motion by controlling sagittal plane motion. A controller, which directly affects side-to-side motion, may provide greater benefits to some individuals. Side-to-side motion can be directly affected by ankle inversion/eversion, which we can control in an ankle-foot prostheses (Panzenbeck and Klute, 2012). By controlling this actuation at each step, some subjects may receive balance restoring assistance.

Overall, in this study, we found that the controller enabled subjects to reduce their intact limb control effort. They also tended to reduce hip actuation effort and overall energy consumption. Three subjects showed a reduction in metabolic rate, one by as much as 23% and changed their preference of the stabilizing controller to a positive one after training. This suggests that this controller is has potential as a balance assistance device.

Chapter 5

Ankle-foot prosthesis emulator with plantarflexion and inversion/eversion actuation

In order to generate novel control ideas that utilize both ankle inversion/eversion and ankle plantar flexion, I developed a two degree-of-freedom ankle-foot prosthesis end-effector with two independently actuated toes. This device is light weight (0.7 kg), while still maintaining exceptional performance. The emulator can generate high power (3 Kw) and high torque (180 Nm in plantarflexion and 30 Nm in inversion/eversion) with good torque tracking (tracking error 5.9 %), high closed-loop torque bandwidth (20 Hz in plantarflexion with 90 Nm amplitude and 24 Hz in in/eversion with 20 Nm amplitude), and high position disturbance rejection band width (18 Hz). These characteristics make it a well-controlled test platform to investigate the effect of ankle actuation controllers on balance-related effort, including step-to-step change in ankle inversion/eversion actuation (Chapter 6 - 7), and a surface probing controller. In addition, the sensory system that contributed to improve performance has been widely used to design other devices including ankle (Witte et al., 2015) and knee exoskeleton.

The contents of this chapter will appear in:

Kim, M., Chen, T., Chen, T., and Collins, S. H., A universal ankle-foot prosthesis emulator with plantarflexion and inversion-eversion torque control for human locomotion experiments, Transactions on Robotics, unpublished.

Collins, S. H., Kim, M., Chen, T., Chen, T. (2015) An ankle-foot prosthesis emulator with control of plantarflexion and inversion-eversion torque. In Proceedings IEEE International Conference on Robotics and Automation, pages 1210-1216*.

* Best Medical Robotics Paper Award, ICRA 2015.

This work is the subject of the following provisional patent:

Kim, M., Chen, T., Chen, T., and Collins, S. H., (2015) An ankle-foot prosthesis emulator with control of plantarflexion and inversion-eversion torque. U.S. Provisional Patent filed May, 2015.

Abstract

Ankle inversion/eversion compliance is an important feature of conventional prosthetic feet, and control of inversion, or roll, in active prostheses could improve balance for people with amputation. We designed a tethered ankle-foot prosthesis with two independently-actuated toes that are coordinated to provide plantarflexion and inversion/eversion torques. A Bowden cable tether provides series elasticity. The prosthesis is simple and lightweight, with a mass of 0.72 kg. Strain gages on the toes measure torque with less than 2% RMS error. Benchtop tests demonstrated a rise time of less than 33 ms, peak torques of 250 N·m in plantarflexion and ±30 N·m in inversion/eversion, and peak power above 3 kW. The phase-limited closed-loop torque bandwidth is 20 Hz with a chirp from 10 to 90 N·m in plantarflexion, and 24 Hz with a chirp from −20 to 20 N·m in inversion. The system has low sensitivity to toe position disturbances at frequencies of up to 18 Hz. Walking trials with an amputee subject demonstrated RMS torque tracking errors of less than 5.1 N·m in plantarflexion and less than 1.5 N·m in inversion/eversion. These properties make the platform suitable for testing inversion-related prosthesis features and controllers in experiments with humans.

5.1 Introduction

Robotic prostheses can improve locomotor performance for individuals who have restricted mobility due to lower-limb amputation. During walking, these devices can restore normal ankle and knee kinematics (Sup et al., 2009), reduce metabolic rate (Herr and Grabowski, 2012), and provide direct neural control of the limb (Huang et al., 2014). As robotic technologies improve, active prostheses are expected to enhance performance even further (Dollar and Herr, 2008; Goldfarb et al., 2013; Cherelle et al., 2014b).

Ankle inversion/eversion, or roll, is an important aspect of prosthesis function.

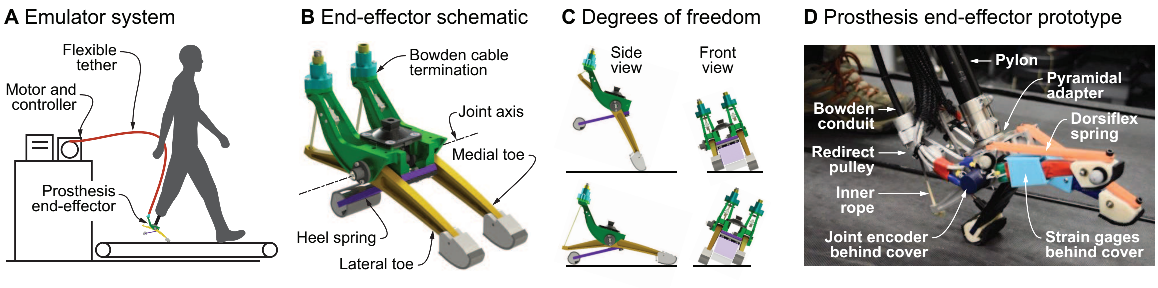

Figure 5.1: Mechanical design of the two degree of freedom ankle-foot prosthesis emulator. **A** The emulator system consists of (1) powerful off-board actuation and control hardware, (2) a flexible Bowden cable tether, and (3) an end-effector worn by the user. **B** The prosthesis end-effector has two independently-actuated toes and a separate, passive heel spring. **C** Plantarflexion occurs when both toes rotate together and inversion/eversion occurs when the medial and lateral toes move in opposite directions. Plantarflexion and inversion/eversion torques are proportional to the sum and difference, respectively, of individual toe torques. **D** The prototype used in experiments is instrumented with encoders at each ankle joint and four strain gages in a Wheatstone bridge on each toe to measure torque. The device is connected to the user via a universal pyramidal adapter. Rubber bands dorsiflex toes during the swing phase of walking.

Commercial prostheses often include a passive inversion/eversion degree of freedom, either using an explicit joint (College Park) or a flexure (Össur). This mitigates undesirable inversion moments created by uneven ground. Inversion moment has a strong effect on side-to-side motions of the body during human walking, and its pattern is altered among individuals with amputation (Hof et al., 2007). Side-to-side motions seem to be less stable in bipedal locomotion (Kuo, 1999; Bruijn et al., 2010), particularly for amputees (IJmker et al., 2014). Difficulty controlling inversion/eversion torque in the prosthetic ankle may partially explain reduced stability (Gates et al., 2013b) and increased fear of falling and fall rates (Miller et al., 2001a) among people with amputation.

Robotic prosthesis designs have begun to incorporate active control of ankle inversion/eversion. Panzenbeck and Klute (Panzenbeck and Klute, 2012) describe a tethered ankle prosthesis with inversion provided by a four-bar linkage and controlled by a linear actuator. The device has a mass of 2.9 kg, can produce torques of up to 34 N·m, and has a 90% rise time of 0.180 s. A plantarflexion degree of freedom is provided using a passive spring. Ficanha et al. (Ficanha et al., 2013) describe a prototype device intended to provide both plantarflexion and inversion/eversion control using two motors and a gimbal joint. The device has a mass of 3.0 kg. Bellman et al. (Bellman et al., 2008) describe a computer model of a similar device, with estimated mass of 2.1 kg. Devices with similar peak torque but lower mass and active control of both plantarflexion and inversion/eversion would enable experimental evaluation of a larger range of assistance techniques.

The mass of prostheses with active inversion/eversion control is related to joint design. Linkages and gimbal joints often involve large parts with complex loading, resulting in increased strength and mass requirements. An alternative is suggested by the split-toe flexures in conventional passive prostheses and the actuation schemes in some powered ankle orthoses (Roy et al., 2009). During walking, peak inversion/eversion torques are of much lower magnitude than peak plantarflexion torques (Eng and Winter, 1995), and the majority of the inversion impulse occurs during periods of high plantarflexion torque (Hunt et al., 2001). Coupling plantarflexion and inversion/eversion torque through the actions of two hinged toes might therefore provide sufficient inversion capacity, allowing a simple, lightweight design.

Mechatronic performance in experimental prosthesis systems can also be improved by separating actuation hardware from worn elements. A tethered emulator approach (Caputo and Collins, 2013, 2014b; Collins, 2013; Collins et al., 2014) decouples the problems of discovering desirable prosthesis functionality from the challenges of developing fully autonomous systems. Powerful off-board motors and controllers are connected to lightweight instrumented end-effectors via flexible tethers, resulting in low worn mass, high torque, high power, and high-fidelity torque control (Caputo and Collins, 2013, 2014b; Witte et al., 2015; Zhang et al., 2015). Such systems can be used to haptically render virtual prostheses to human users, facilitating the discovery of novel device behaviors (Kim and Collins, 2015b; Jackson and Collins, 2015) that can then be embedded in separate autonomous designs (Collins et al., 2015b). This approach can also be used for rapid comparison of commercial prostheses in a clinical setting (Collins et al., 2014; Caputo et al., 2015a). To be most effective, such prosthesis emulators should have high closed-loop torque bandwidth and lightweight, strong, accurately-instrumented end-effectors.

Series elasticity can have a strong effect on the quality of torque control in a robotic emulator system. Adding a spring in series with a high-stiffness transmission can reduce sensitivity to unexpected actuator displacements (Pratt and Williamson, 1995) such as those imposed by the human. Unfortunately, series compliance also reduces force bandwidth when the output is fixed, because the motor must displace further to stretch the spring. The optimal stiffness strikes a balance between these competing factors for a particular system and task. In a tethered emulator, the flexible transmission itself may have significant compliance, which might provide appropriate series elasticity.

Here we describe the design and evaluation of a robotic ankle-foot prosthesis emulator system with active control of both plantarflexion and inversion/eversion torques. We designed an end-effector that allowed inversion/eversion using two articulated toes, which we aimed to make lightweight and strong. We integrated the end-effector with existing off-board motor and control hardware, expected to facilitate high-bandwidth torque control. The end-effector did not include explicit series elasticity, testing the sufficiency of axial compliance in the tether. We implemented a basic walking controller, intended to evaluate the system's potential for emulating prosthesis behavior during interactions with a human user. We expect this approach to result in validation of a system that can explore new dimensions of prosthesis assistance, particularly those related to balance during walking.

An earlier version of this work was presented at the *International Conference on*

Robotics and Automation (Collins et al., 2015a). In this paper, we present the results of additional benchtop tests of peak torque and peak power, the results of additional walking trials with a subject with transtibial amputation, expanded methods, results and discussion, and supplementary videos.

5.2 Methods

We designed and constructed an ankle-foot prosthesis end-effector with torque control in both plantarflexion and inversion/eversion directions. We characterized system performance in benchtop tests, including peak torque, peak power, torque control bandwidth and disturbance rejection, and characterized torque tracking performance during walking under a variety of conditions.

5.2.1 Mechanical Design

The two degree of freedom ankle-foot prosthesis was designed as an end-effector for a tethered emulator system (Fig. 5.1A). Powerful actuation and control hardware is located off-board so as to keep worn mass low. Flexible Bowden cable tethers transmit mechanical power to the prosthesis, but do not interfere with natural movements of the limb. We used two 1.61 kW AC servomotors with 5:1 planetary gearheads and dedicated motor drives (Baldor Electric Corp., Fort Smith, AR), controlled by a 1 GHz real-time controller (dSPACE Inc., Wixom, MI). Bowden cables comprised coiled-steel outer conduits (Lexco Cable Mfg., Norridge, IL) and 3 mm synthetic inner ropes (Vectran Fiber Inc., Fort Mill, SC). The motor, real-time controller and tether are described in detail in (Caputo and Collins, 2014b).

The ankle-foot prosthesis achieves torque and motion in both plantarflexion and inversion/eversion directions using two independent toes. The toes share a single axis of rotation similar to the plantarflexion axis in the human ankle joint, and are spaced mediolaterally such that one is closer to the centerline of the body

(Fig. 5.1B). Plantarflexion occurs when both toes rotate in the same direction, and inversion/eversion occurs when they rotate in opposite directions (Fig. 5.1C; Video 1). We define plantarflexion angle as the average of the toe angles and inversion/eversion angle as the difference between toe angles multiplied by the ratio of toe length to half the foot width. Similarly, plantarflexion torque, τ_{pf}, is defined as the sum of the lateral and medial toe torques, τ_l and τ_m, while inversion torque, τ_{inv}, is defined as the difference between the lateral and medial toe torques multiplied by the ratio of half the foot width, $\frac{1}{2}w$, to toe length, l, or

$$\begin{aligned} \tau_{pf} &= \tau_l + \tau_m \\ \tau_{inv} &= \tfrac{w}{2l}(\tau_l - \tau_m) \end{aligned} \tag{5.1}$$

Toes are actuated through independent Bowden cable tethers and off-board motors, allowing independent control of medial and lateral toes. Plantarflexion and inversion/eversion torques can be independently controlled, but maximum allowable inversion/eversion torque is proportional to plantarflexion torque. When inversion torque is zero, the plantarflexion torque is divided evenly between the toes. As inversion torque increases towards its limit, the torque on the lateral toe approaches the total desired plantarflexion torque, while the torque on the medial toe approaches zero. When inversion (or eversion) torque equals plantarflexion torque divided by the ratio of toe length to half the foot width, $\frac{2l}{w}$, the inversion/eversion torque cannot be increased further, since doing so would require negative torque on the medial (or lateral) toe, and negative ground reaction forces. This defines a feasible region of inversion torques as a function of plantarflexion torque: $|\tau_{inv}| \leq \frac{w}{2l}\tau_{pf}$. For torque patterns typical of human walking, inversion/eversion torques lie within the feasible region during most of stance (Fig. 5.2).

The prosthesis consists of a frame, two toes with revolute joints, and a compliant heel. The frame of the device (Fig. 5.1D) is connected to the user's pylon or socket via a universal pyramidal adapter (Video 1). The frame houses needle roller bearings for the ankle joints, which have a double-shear construction. Each toe is long and thin, tapers towards its ends, and has an I-beam cross section, making it well-suited to three-point bending. One end of the toe contacts the ground, while the other end is acted on by the Bowden cable, with the hinge located in the middle. When the inner rope of the Bowden cable pulls upwards on the posterior aspect of the toe, a moment is generated. The Bowden cable conduit presses down on the frame equally and oppositely, such that the foot experiences no net force from the transmission. Rubber bands act to dorsiflex the toe when the transmission allows, such as during the swing phase. A separate, unactuated heel spring is connected to the frame. Rubber-coated plastic pads are attached to the ends of the heel and toes to improve traction against the ground. The frame and toes were machined from 7075-T6 aluminum, the heel spring was machined from fiberglass (GC-67-UB, Gordon Composites, Montrose, CO, USA), and the toe pads were fabricated using fused-deposition modeling of ABS plastic. CAD models and a bill of materials are located at (Kim et al.).

Prosthesis dimensions were based on an average male human foot (Hawes and Sovak, 1994). The device measures 0.23 m in length, heel to toe, 0.07 m in width, toe center to toe center, and 0.08 m in height, from ground to ankle joint. The toe length, from axis of rotation to tip, is 0.14 m. Ankle range of motion is -20° to 30° in plantarflexion and greater than -30° to 30° in inversion/eversion, greater than the range observed during normal walking (Winter, 1991) and comparable to the range of the human ankle joint (Roaas and Andersson, 1982). The prosthesis end-effector weighs 0.72 kg.

The end-effector did not include an explicit spring in the transmission, but some series elasticity was provided by the Bowden cable. Series elasticity can improve torque tracking in the presence of disturbances from the human user (Vallery et al., 2008). In our prior designs (Caputo and Collins, 2013; Witte et al., 2015), we used

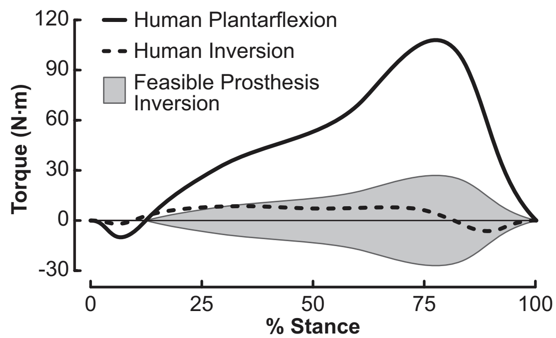

Figure 5.2: Coupling between prosthesis plantarflexion and inversion/eversion torque illustrated with typical human walking data. Maximum feasible inversion/eversion torque (gray region) is proportional to plantarflexion torque (Eq. 5.1). With a typical plantarflexion torque pattern (solid line) the typical inversion/eversion torque (dashed line) falls within the feasible region for this device. Reference data for human walking at 1.6 m·s^{-1} are from Hunt et al. (2001).

fiberglass leaf springs or steel coil springs at the connection between the Bowden cable and the hinged foot element, resulting in combined rotational stiffnesses of between 140 and 320 N·m·rad^{-1}. In this design, we explored whether the compliance of the Bowden cable itself might be sufficient to facilitate low-error torque tracking. This would have the benefit of reducing the mass and complexity of the end-effector. In tests where the off-board motors were fixed while the prosthesis toes were rotated, we measured an effective stiffness of about 550 N·m·rad^{-1}. With increased series stiffness, we expected joint torque to change more quickly when toes were fixed and the motor was rotated, resulting in higher closed-loop torque bandwidth. However, we also expected torques to change more quickly when the motor was stationary and the toes were unexpectedly rotated, for example during initial contact with the ground, which could result in poorer torque tracking under realistic conditions. We therefore separately tested bandwidth, disturbance rejection and torque tracking during walking, as described below.

Medial and lateral toe joint angles were sensed individually using digital absolute magnetic encoders (MAE3, US Digital, Vancouver, WA). Toe torques were sensed

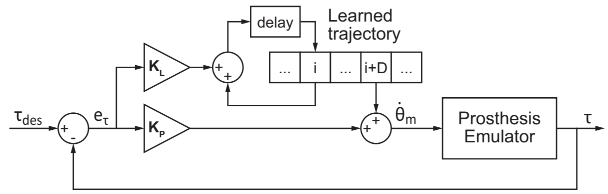

Figure 5.3: Block diagram illustrating the torque control approach. Desired torque, τ_{des}, is compared to measured torque, τ, to obtain torque error, e_{τ}. In the feedback loop, a proportional gain, $\mathrm{K_P}$, is applied to the error and used to set desired motor velocity, $\dot{\theta}_m$. The feed-forward compensation used during walking trials is updated by applying a learning gain, $\mathrm{K_L}$, to the torque error and adding the result to the existing value of the learned trajectory of motor velocity commands for this instant in time, i. The update takes effect on the next walking step. The previously-learned compensation is used to command desired motor velocity on this walking step, adding to the feedback loop. The feed-forward compensation value is from an instant D control-loop cycles in the future, reflecting an anticipated delay in achieving the desired motor velocity after it is commanded.

using strain gages (SGD-3, Omega Engineering, Stamford, CT), configured in a Wheatstone bridge, with two gages on the top and bottom surfaces of each toe midway between the tip and the ankle joint. Heel contact was sensed using strain gages on the heel spring, with a half bridge configuration (KFH-6, Omega Engineering). Bridge voltage was amplified (FSH01449, Futek, Irvine, CA), sampled at a frequency of 5000 Hz and low-pass filtered with a cutoff frequency of 100 Hz. Plantarflexion and inversion/eversion angles and torques were calculated in software from medial and lateral toe values.

5.2.2 Control

We used classical feedback control to regulate torque during benchtop tests, with an additional iterative learning term during walking trials (Fig. 5.3). Desired torque for each toe was first calculated from desired plantarflexion and inversion/eversion torques. Motor velocities were then commanded using proportional control on toe torque error. Motor velocity is similar to the rate of change in toe torque, owing to compliance between the off-board motor and prosthesis toe. During walking trials, an additional time-based iterative learning term was added, which provided feed-forward compensation of torque errors that tended to occur at the same time each step. This method is described in detail in (Zhang et al., 2015).

In walking trials, torque control was used during stance and position control was used during swing. Initial toe contact was sensed from an increase in toe torque upon making contact with the ground. During the ensuing stance period, desired inversion/eversion torque was set to a constant value, providing a simple demonstration of platform capabilities. Desired plantarflexion torque during stance was calculated as a function of plantarflexion angle, as described in (Caputo and Collins, 2014a), so as to approximate the torque-angle relationship observed at the ankle during normal walking (Winter, 1991). Toe off was detected when plantarflexion torque crossed a minimum threshold. During the ensuing swing phase, toes were position controlled to provide ground clearance.

5.2.3 Experimental Methods

We conducted benchtop tests to characterize device performance in terms of torque measurement accuracy, response time, bandwidth, peak torque, peak power and disturbance rejection. We performed walking trials to assess mechatronic performance under similar conditions as expected during biomechanics experiments.

C.1 Benchtop Testing Methods

Torque measurement calibration was performed by applying known forces to the end of each toe using free weights and fitting amplified strain gage bridge voltage to applied torque. Measurement accuracy was characterized in a separate validation test as root mean squared (RMS) error between applied and measured toe torques.

Step response tests were performed in which we rigidly fixed the prosthesis frame and toes and commanded desired torque as a square wave from 0 to 180 N·m in plantarflexion or -20 to 20 N·m in inversion/eversion. We conducted 10 trials for each direction and computed the mean and standard deviation of the 90% rise and fall

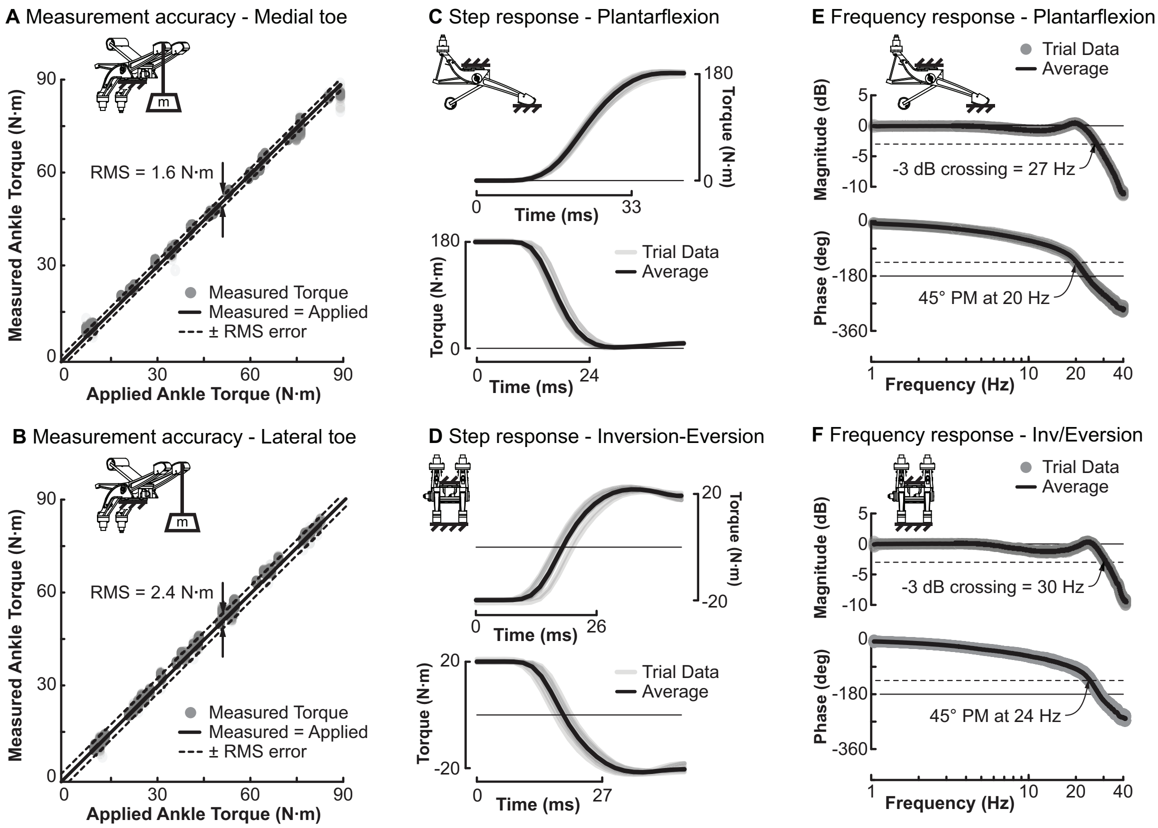

Figure 5.4: Benchtop tests with a fixed load demonstrate low torque measurement error, fast rise time and high closed-loop torque bandwidth in both plantarflexion and inversion/eversion directions. Torque measurement validation for the **A** medial and **B** lateral toes. Step responses for closed-loop control of **C** plantarflexion and **D** inversion/eversion torque. Rise and fall times ranged from 0.024 to 0.033 s. Bode plots for closed-loop control of **E** plantarflexion and **F** inversion/eversion torque, calculated from responses to 90 N·m and ±20 N·m magnitude chirps in desired torque, respectively. Bandwidth ranged from 20 to 30 Hz, limited by the 45° phase margin criterion.

times.

We performed bandwidth tests in which desired torque was commanded as a 0 to 40 Hz chirp, oscillating between 10 and 90 N·m for plantarflexion and between -20 and 20 N·m for inversion/eversion. We used an exponential chirp to improve signal to noise ratio in the low frequency range. We transformed the desired and measured torque into the frequency domain using a Fast Fourier Transform and used the magnitude ratio and phase difference to generate a Bode plot. We calculated the gain-limited and phase-limited bandwidths (Warwick, 1996) as the frequencies at

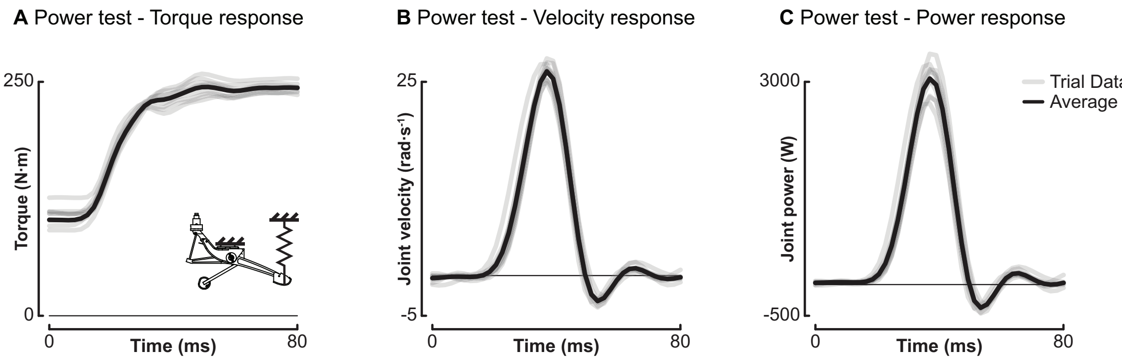

Figure 5.5: Benchtop tests with a compliant load demonstrate high peak torque, velocity and power at the ankle joint. **A** Measured plantarflexion torque peaked at about 250 N·m. **B** Measured plantarflexion velocity peaked at above 25 rad·s^{-1}. **B** Measured joint mechanical power peaked at about 3 kW.

which the amplitude ratio was -3 dB and the phase margin was 45°, respectively. We performed 10 trials for both torques and calculated crossover frequency means and standard deviations.

Peak torque and peak power were characterized using step responses with a compliant load. We rigidly fixed the prosthesis frame to the benchtop and attached the toes to the benchtop through a coil spring with stiffness of 63,000 N·m^{-1}. We then commanded desired plantarflexion torque as a step increase from about 100 to 250 N·m. We conducted 10 trials and computed the mean and standard deviation of the peak torque and peak power for each trial.

We also performed a test intended to evaluate the torque errors that would arise from unexpected disturbances to toe position. We expected that high series stiffness in this system might have provided high bandwidth at the cost of higher sensitivity to position disturbances, for example during initial toe contact with the ground. We placed the toes on opposite ends of a seesaw-like testing jig such that toe forces were equal and toe motions were equal and opposite (Video 2). We then applied a 0 to 25 Hz chirp in medial toe position, oscillating between 0° and 5° of plantarflexion (or 0 and 0.012 m of toe tip displacement) and commanded a constant desired torque of 30 N·m to the lateral toe. We transformed the amplitude of the resulting torque error

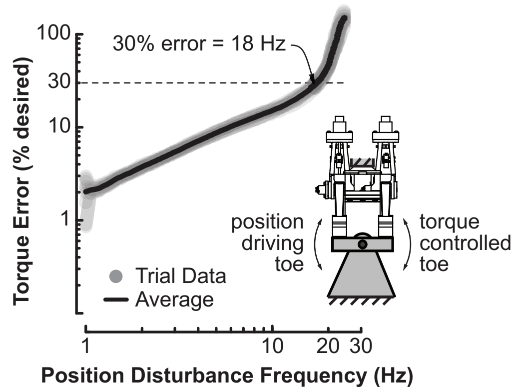

Figure 5.6: Disturbance rejection, depicted as the relationship between torque error (% of the constant desired value) versus the frequency of an applied disturbance in toe position. This characterizes the ability of the system to reject unexpected environmental disturbances, such as from sudden contact with the ground. Torque error was less than 30% of the desired value of 30 N·m for disturbance frequencies up to 18 Hz.

into the frequency domain using a Fast Fourier Transform, reported as a percent of the constant desired torque magnitude. We calculated the frequency at which error rose above 30% of the desired torque, analogous to the -3 dB (70% amplitude) criteria used in bandwidth tests.

C.2 Walking Demonstration Methods

We performed two sets of walking trials to evaluate torque tracking performance under realistic conditions. In the first set of trials, one subject (67 kg, 1.77 m tall, 23 yrs, male) without amputation wore the device using a simulator boot (Caputo and Collins, 2014a). We used minimal Bowden cables, about 2 m in length, for best torque tracking performance. Five walking trials were conducted in which desired inversion/eversion torque was commanded as: Maximum, 15 N·m, 0 N·m, -15 N·m, and Maximum Negative. The magnitudes of Maximum and Maximum Negative inversion torque were proportional to plantarflexion torque at each instant in time. Using the simulator boot allowed these large torques to be applied comfortably.

In the second set of trials, one subject with unilateral transtibial amputation (89 kg, 1.72 m tall, 26 yrs, male) wore the device using their prescribed socket. We

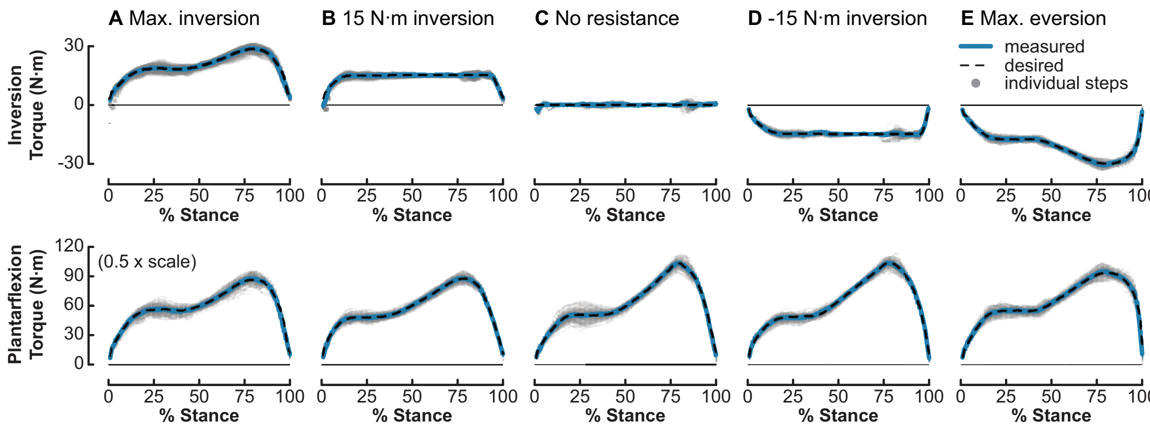

Figure 5.7: Torque tracking during walking experiments. Desired ankle inversion torque was set to **A** Maximum, **B** 15 N·m, **C** zero, **D** -15 N·m, and **E** Maximum Negative, while desired plantarflexion torque was a consistent function of ankle plantarflexion angle. Maximum and Maximum Negative allowable inversion torque were limited by desired plantarflexion torque, since toe ground reaction forces could not become negative. In each 100-stride trial, measured torque closely matched desired torque, with RMS errors of at most 3.7 N·m in plantarflexion and 1.1 N·m in inversion/eversion across conditions. Differences between average torque and individual-step torques were dominated by changes in desired torque arising from natural variability in the subject's gait pattern.

used extended Bowden cables, about 4 m in length, for best range of movement on the treadmill. Three walking trials were conducted in which desired inversion/eversion torque was commanded as: 10 N·m, 0 N·m and -10 N·m. These magnitudes were chosen to maximize range of torque without causing discomfort in the residual limb from repeated applications of torque in one direction (Video 3).

In both sets of trials, subjects walked on a treadmill at 1.25 m·s^{-1} for 100 steady-state strides in each condition. We normalized each step to percent stance period and calculated an average step for each condition. We characterized torque tracking error as both the RMS error across the entire trial and as the RMS error of the average step. We did not measure human biomechanical response, since this study was intended to evaluate performance of the robotic system and not the effects of a proposed intervention.

5.3 Results

Benchtop tests with a fixed load revealed low torque measurement error, fast rise time and high closed-loop torque bandwidth. The root mean squared (RMS) torque measurement errors for medial and lateral toes were 1.64 N·m and 2.43 N·m, respectively, following calibration (Fig. 5.4A&B). The 90% rise and fall times between 0 and 180 N·m in plantarflexion torque were 0.033 ± 0.001 s and 0.024 ± 0.001 s (mean $\pm$ s.d.), with 0.5% and 1.6% overshoot, respectively (Fig. 5.4C). The 90% rise and fall times between -20 to 20 N·m in inversion/eversion torque were 0.026 ± 0.002 s and 0.027 ± 0.002 s, with 3.0% and 3.2% overshoot, respectively (Fig. 5.4D). With desired plantarflexion torque oscillating between 10 and 90 N·m, the -3 dB magnitude and 45° phase margin crossover frequencies were 27.2 ± 0.2 Hz and 20.3 ± 0.3 Hz, respectively (Fig. 5.4E). With desired inversion/eversion torque oscillating between -20 and 20 N·m, the -3 dB magnitude and 45° phase margin crossover frequencies were 29.8 ± 0.2 Hz and 23.8 ± 0.3 Hz, respectively (Fig. 5.4F).

Benchtop tests with a compliant load revealed high peak joint torque, velocity and power. Peak measured plantarflexion torque was 248 ± 6 N·m (Fig. 5.5A). Peak measured plantarflexion velocity was 26.3 ± 1.1 rad·s^{-1} (Fig. 5.5B). Peak mechanical power was 3,050 $\pm$ 240 W (Fig. 5.5C). During the period of peak power output the tether was being stretched, thereby absorbing energy and not contributing to peak power through return of stored energy.

When we applied a 0.012 m amplitude chirp disturbance in toe endpoint position and commanded a constant desired torque of 30 N·m, torque error was less than 30% up to a disturbance frequency of 18 Hz (Fig. 5.6). This disturbance frequency and amplitude are similar to unexpected contact with stiff ground at a rate of 1.4 m·s^{-1}.

In the first set of walking trials, the non-amputee subject walked comfortably with the prosthesis on a short tether while five levels of constant desired

Table 5.1: Torque tracking errors during 100 steps of walking with various values of desired inversion/eversion torque.

Case 1: Non-amputee with simulator boot and short tether

	Plantarflexion Torque Tracking				*Inversion/Eversion Torque Tracking*			
Inv/eversion torque	RMS error	$\%\tau_{max}$	Avg RMS error	$\%\tau_{max}$	RMS error	$\%\tau_{max}$	Avg RMS error	$\%\tau_{max}$
τ_{inv} = Max.	3.2 ± 1.1 N·m	3.7%	1.3 N·m	1.6%	1.1 ± 0.4 N·m	3.8%	0.4 N·m	1.6%
τ_{inv} = -15 N·m	1.9 ± 0.4 N·m	2.2%	0.7 N·m	0.8%	0.9 ± 0.2 N·m	5.9%	0.7 N·m	4.4%
τ_{inv} = 0	2.9 ± 1.7 N·m	2.8%	0.6 N·m	0.6%	0.8 ± 0.2 N·m	-	0.5 N·m	-
τ_{inv} = -15 N·m	2.9 ± 0.8 N·m	2.8%	0.9 N·m	0.8%	0.8 ± 0.2 N·m	5.6%	0.3 N·m	2.1%
τ_{inv} = Neg. Max.	3.0 ± 0.9 N·m	3.3%	1.3 N·m	1.4%	1.0 ± 0.3 N·m	3.3%	0.4 N·m	1.6%

Case 2: Transtibial amputee with long tether

	Plantarflexion Torque Tracking				*Inversion/Eversion Torque Tracking*			
Inv/eversion torque	RMS error	$\%\tau_{max}$	Avg RMS error	$\%\tau_{max}$	RMS error	$\%\tau_{max}$	Avg RMS error	$\%\tau_{max}$
τ_{inv} = -10 N·m	4.7 ± 1.0 N·m	3.9%	1.3 N·m	1.1%	1.5 ± 0.3 N·m	14.6%	0.7 N·m	6.4%
τ_{inv} = 0	5.1 ± 1.2 N·m	4.2%	1.5 N·m	1.3%	1.1 ± 0.3 N·m	-	0.2 N·m	-
τ_{inv} = -10 N·m	4.8 ± 1.0 N·m	3.9%	1.2 N·m	1.0%	1.3 ± 0.3 N·m	13.2%	0.3 N·m	2.5%

inversion/eversion torque were applied (Fig. 5.7). Peak inversion torques during Maximum and Maximum Negative conditions were about 30 N·m and −30 N·m, respectively. Torque tracking errors in both plantarflexion and inversion/eversion directions were low, with maximum RMS errors of 3.2 N·m (3.7% of peak) in plantarflexion torque and 1.1 N·m (3.8% of peak) in inversion/eversion torque, across all conditions (Table 5.1).

In the second set of walking trials, the transtibial amputee subject walked comfortably with the prosthesis on the longer tether while three lower levels of constant desired inversion/eversion torque were applied. Torque tracking errors in both plantarflexion and inversion/eversion directions were about slightly higher in these trials, with maximum RMS errors of 5.1 N·m (4.2% of peak) in plantarflexion torque and 1.5 N·m (14.6% of peak) in inversion/eversion torque (Table 5.1). Higher percent error in inversion/eversion torque in this set of trials was primarily the result of lower peak torque (±10 N·m vs. ±30 N·m).

5.4 Discussion

We designed, built and tested an ankle-foot prosthesis system with torque control in both plantarflexion and inversion/eversion directions. The end-effector is lightweight, yet provides high torque, speed and power. The system has both high closed-loop torque bandwidth and low torque errors in the presence of unexpected toe displacements. During walking trials, a wide range of inversion/eversion torque values were tracked with low error. Taken as a whole, these results demonstrate the versatility of the ankle-foot prosthesis emulator and its suitability for haptic emulation of prostheses with both pitch and roll degrees of freedom.

This prosthesis emulator is versatile, with mass, size, torque, speed and power that compare favorably to normal ankle-foot function and to other active prostheses. The end-effector has about 60% of the mass of a typical human foot (Winter, 1990), similar to the mass of passive ankle-foot prostheses (Össur) and about a third of the mass of other tethered (Panzenbeck and Klute, 2012; Ficanha et al., 2013; Huang et al., 2014) and untethered (Hitt et al., 2010; Herr and Grabowski, 2012; Shultz et al., 2013; Cherelle et al., 2014a) powered ankle-foot prostheses. The end-effector has dimensions similar to a human foot (Hawes and Sovak, 1994). Peak measured plantarflexion and inversion/eversion torques were 50% and 230% greater, respectively, than the peak values observed at the human ankle joint during walking and running among typical males (Whittle, 1996; Hunt et al., 2001; Novacheck, 1998). Peak measured plantarflexion torques were about 40% greater than in other devices with powered plantarflexion (Hitt et al., 2010; Herr and Grabowski, 2012; Shultz et al., 2013; Huang et al., 2014; Cherelle et al., 2014a), and peak inversion/eversion torques were equivalent to those in other devices with powered inversion/eversion (Panzenbeck and Klute, 2012). Peak joint velocity and power were each about three times greater than peak values observed at the ankle joint during normal walking and running (Whittle, 1996; Novacheck, 1998), and an order of magnitude greater than in previous powered devices (Hitt et al., 2010; Au and Herr, 2009; Shultz et al., 2013; Huang et al., 2014; Cherelle et al., 2014a).

The responsiveness of this device also compares favorably to human musculature and to other active prostheses, allowing accurate rendering of virtual devices. The system has high closed-loop torque bandwidth, a limiting factor in the fidelity of haptic emulation (Abul-Haj and Hogan, 1987; Hannaford and Okamura, 2008; Griffiths et al., 2011). Measured bandwidth was about four times that of human ankle muscles (Bawa and Stein, 1976). This is nearly twice the bandwidth of our previous ankle-foot prosthesis system (Caputo and Collins, 2014b), five times that of untethered electric prostheses (Au et al., 2007), and about ten times that of similar systems using pneumatic muscles (Huang et al., 2014; Gordon et al., 2006). Inversion/eversion step response time was about five times faster than prior systems with on-board actuation (Panzenbeck and Klute, 2012). Torque disturbances due to unexpected toe movements could be rejected at high frequencies, an indication of robustness during unpredictable human interactions (Hogan, 1984). Torque tracking errors were below 30% in the presence of disturbances at up to twice the peak voluntary oscillation frequencies of the human ankle (Agarwal and Gottlieb, 1977). This disturbance rejection cutoff frequency corresponded to more than 83% of the frequency content of the prosthetic ankle joint angle during walking trials.

Both plantarflexion and inversion/eversion torques were tracked with low error during walking across a range of conditions, demonstrating the effectiveness of this system for prosthesis emulation experiments. Absolute torque tracking errors were low across all conditions and outcomes, with values similar to those observed for humans attempting to maintain constant isometric ankle joint torque (Vuillerme and Boisgontier, 2008). Maximum observed plantarflexion and inversion/eversion torque errors were 2% and 5% of system torque capacity, respectively. In most cases errors were also small relative to peak desired torques, although percent error naturally approached infinity as desired inversion/eversion torque approached zero.

Both absolute and relative torque errors were greater in tests with the longer tether and the amputee subject. Absolute tracking errors were about 50% higher in both plantarflexion and inversion/eversion, likely due to increased compliance, friction, stiction, delays and other nonlinearities with the longer Bowden cable. This decrease in absolute performance could also relate to differences between amputee and non-amputee gait characteristics, but such differences were not apparent in any measured kinetics or kinematics data. Use of shorter, straighter Bowden cables is therefore warranted where possible, for example by mounting motors above the subject (Andersen and Sinkjær, 1995). Other improvements to the Bowden cable transmission, for example using intermediate components with lower friction and fewer nonlinearities, could yield simultaneous improvements in torque tracking, range on the treadmill, and convenience. The substantially higher percent inversion/eversion torque error observed in trials with the amputee subject are largely the result of lower desired torques. When maximum inversion/eversion torques were applied on each step, the subject reported discomfort in their residual limb. It is therefore not clear whether the full range of inversion/eversion torque capacity of the present system is necessary for tests involving subjects with amputation. Intermittent application of higher torques may be allowable, and peak torques may vary across individuals.

Although this design does not include an explicit series spring in the end-effector, disturbance rejection was relatively high and torque tracking errors were low during walking. It appears that series elasticity provided by stretch in the Bowden cable transmission sufficiently decoupled the toes from the inertia of the motor. This has not been the case for all emulator end-effectors we have tested. In pilot tests with an ankle exoskeleton (Zhang et al., 2015), we found that removing the coil spring at the ankle joint led to increased torque tracking errors. Differences may be related to the types of disturbance provided by the human in these cases; having muscles in parallel with the actuator, as with an exoskeleton, may produce larger or higher-frequency variations in interaction torques than when a prosthesis is placed in series with the limb.

Torque measurement was also not adversely affected by lack of a series spring in this system. Measuring torque using spring deflection (Caputo and Collins, 2013; Pratt et al., 2002) can reduce electromagnetic noise compared to strain gages (Pratt and Williamson, 1995). In this case, the amplified strain gage bridge voltage exhibited noise in the kHz range, but this was easily removed by sampling at high frequency and low-pass filtering. Utilizing Bowden cable compliance therefore reduced the mass and complexity of the end-effector without negatively affecting torque tracking or measurement.

Instrumenting the toes with strain gages also resulted in lower complexity and more accurate torque measurement than the use of load cells in this case. In an earlier revision of this design, Bowden cable tension was sensed using pushbutton load cells with a through hole at the conduit termination (inside the cyan elements in Fig. 5.1B). This resulted greater mass, parasitic loads from the cable, and hysteresis due to friction and shifting at the termination.

Using two toes for inversion/eversion results in a simple, lightweight structure, but does not provide direct measurement of frontal-plane motions. The angle of the shank with respect to vertical in the frontal plane cannot be calculated from the angles of the medial and lateral toes alone (unless they are equal), since rotation about the line between toe contact points is not captured by joint angles. More sensory information, such as the pitch angle of the prosthesis frame, is required. A similar problem arises if inversion/eversion torque is defined about an axis in the direction of travel. In a laboratory setting, this issue can be overcome by measuring shank angle directly with motion capture equipment. Solutions that would be suitable for autonomous devices include measuring shank angle with an inertial measurement unit or (actively) maintaining heel contact throughout stance to obtain the missing configuration-related measurement.

The prosthesis emulator has high-fidelity control over the medial-lateral location of the center of pressure during stance, but would require an additional active degree of freedom to usefully control fore-aft center of pressure location. Humans seem to regulate the path of the center of pressure during walking (Hansen et al., 2004), making this a potentially interesting signal for manipulation. In this system, the medial-lateral center of pressure can be controlled through inversion/eversion torque when both toes are in contact with the ground. In the fore-aft direction, the center of pressure can only be controlled when the heel is also in contact. Since the heel is passive, contact is maintained only for a limited range of shank and toe configurations. Active torque control of the heel would resolve this issue.

Although we only present data for tests with two subjects, we expect similar haptic emulation performance for a wide range of individuals and protocols. Human response to robotic intervention can depend strongly upon subject characteristics (Zelik et al., 2011; Segal et al., 2012), but device behavior typically does not (Major et al., 2011; Adamczyk et al., 2013). Benchtop measurements are, of course, subject-independent. This study concerned the mechatronic performance of the prosthesis emulator, whereas future studies probing biomechanical response to different interventions will require multiple subjects with amputation.

This system provides exceptional versatility within a laboratory environment, but cannot be used for community ambulation. This is a fundamental limitation of the approach compared to mobile devices. One implication is that acclimation to use of the device must take place in the laboratory, which places a practical limit on training time. Positive outcomes with some active prostheses have been achieved following several weeks of acclimation (Herr and Grabowski, 2012), although adequate adaptation times are not yet known. Use of a subject's prescribed prosthesis between training sessions could also cause interference effects, like those observed during manipulation of novel objects (Fu and Santello, 2015). Some aspects of the dynamics of treadmill walking differ from those of overground walking (Dingwell et al., 2001), which could limit the applicability of some findings to community ambulation. For experimental protocols exploring the design and control of novel prostheses in a laboratory setting, however, this system provides exceptional performance.

5.5 Conclusions

We have described the design of a tethered ankle-foot prosthesis emulator system with independent control over plantarflexion and inversion/eversion torque. Benchtop tests and experiments during human walking provided a detailed characterization of system dynamics and performance, which we expect will guide the design of improved systems. The torque control fidelity of this platform was exceptional, particularly in terms of closed-loop torque bandwidth, making it suitable for haptic emulation of prostheses with pitch and roll degrees of freedom. A wide variety of virtual devices could be rendered to users as part of the clinical prescription process (Collins et al., 2014; Caputo et al., 2015a), during the development of new commercial devices (Collins, 2013), or in basic science experiments probing the nature of human locomotion (Caputo and Collins, 2014a; Kim and Collins, 2015b). In particular, we expect experiments with this system to provide insights into the role of inversion/eversion torque on walking balance for individuals with amputation.

5.6 Acknowledgments

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Chapter 6

The ankle inversion/eversion stiffness influences balance to unilateral, transtibial amputees

In this study, an ankle inversion/eversion controller was designed. The controller restores torque on each step by multiplying stiffness by a deviation of ankle inversion/eversion angle from a nominal value during stance, while keeping average ankle inversion/eversion torque zero. I then conducted an experiment to test this control strategy on individuals with below knee amputation (N = 4). For this experiment, I provided stabilizing controllers and destabilizing controllers. Stabilizing controllers delivered restoring torque during stance and destabilizing controllers acted oppositely. I found the stabilizing controllers can improve balance, as shown by reduced foot placement control effort, intact limb control effort, and average step width, as well as favorable subject-satisfaction rating when compared to the destabilizing controller. The stabilizing controllers, however, failed to reduce metabolic rate. These results suggest that ankle inversion/eversion stiffness can affect balance but, in order to reduce overall energy consumption, additional training or modifications to the controller might be necessary. Perhaps, by modulating inversion/eversion torque on each step and providing a forced exploration period, we may be able to reduce metabolic cost, in addition to other balance-related measures (Chapter 7).

The contents of this chapter will appear in:

Kim, M., Lynees H, Chen, T., Collins, S. H. (2015) The Influence of Inversion/Eversion Stiffness of an Ankle-Foot Prostheses on Balance-Related Effort during Walking, **in preparation**.

A preliminary version of this work was presented at:

Kim, M., Chen, T., Lynees H, Collins, S. H.(2015) The Influence of Inversion-Eversion Stiffness of Ankle-Foot Prosthesis on Amputee Balance-Related Effort during Walking. Poster presentation at Dynamic Walking.

Abstract

Prosthesis features that enhance balance are desirable to people with below-knee amputation. Ankle inversion/eversion compliance is intended to improve balance on uneven ground, but its effects remain unclear even on level ground. We posited that increasing ankle inversion stiffness during level-ground walking would reduce balance-related effort by assisting in recovery from small disturbances in frontal-plane motions. We performed tests with an ankle-foot prosthesis emulator programmed to apply inversion torques in proportion to the deviation from a nominal inversion angle trajectory. We applied a range of stiffnesses, hypothesizing that positive stiffnesses would reduce effort while negative stiffnesses would increase effort. Nominal trajectories were calculated online as a moving average over several steps. In experiments with K3 ambulators with unilateral transtibial amputation (N = 5), stiffness affected step width variability, average step width, intact-foot center of pressure variability, and user satisfaction ($p \leq 0.005$, ANOVA), but not metabolic rate ($p = 0.4$). High positive stiffness reduced step width variability by 28%, reduced step width by 12%, reduced center of pressure variability by 29%, and increased satisfaction by 81% ($p \leq 0.03$, paired t-tests) compared to high negative stiffness. Ankle inversion stiffness can have a substantial effect on balance-related effort.

6.1 Introduction

Below knee amputation is the most common type of major amputation worldwide (Lerner and Soudry, 2011). In the United States alone, there are more than a quarter million individuals with this class of amputation, and this number is predicted to more than double by 2015 (Ziegler-Graham et al., 2008). One of the main consequences of below knee amputation is a loss of balance and balance confidence (Miller et al., 2001a, 2002). There is also a notable desire among lower-limb amputees for a prosthetic device that provides balance assistance (Legro et al., 1999). Improved ankle-foot prosthesis technology might help meet these needs.

Passive compliance is a common feature in ankle-foot prostheses that could be used to improve balance (Lindhe, 2014; Kim et al., 2007). Based on previous research, prostheses with ankle compliance seem to decrease walking effort compared to rigid devices by accommodating uneven ground conditions (Childers et al., 2015). Surprisingly, this study also found that a similar level of compliance seemed to be beneficial during level ground conditions. These unexpected results may be due to an intrinsic coupling of sagittal and frontal plane compliance in commercial devices. In the sagittal plane, compliance may play a role in navigating uneven terrain (Gates et al., 2013b), but it may be more strongly tied to natural transition of the center of pressure location, related to gait speed (Hansen et al., 2004). In the frontal plane, however, a stiffer device may be favorable on level ground, because it can provide a restoring torque that pushes the body towards a more upright position. Perhaps by separately examining frontal plane and sagittal plane compliance, the effects on balance might be more clearly understood.

Maintenance of balance in the frontal plane seems to require active effort during human walking (Kuo, 1999; Collins and Kuo, 2013), especially for individuals with transtibial amputation (Beltran et al., 2014; Gates et al., 2013b). Careful manipulation of ankle inversion and eversion torque in an ankle-foot prosthesis may be able to increase balance in the frontal plane (Kim and Collins, 2013; Kuo, 1999). To test the effect of ankle inversion on balance, researchers have recently focused on developing devices with powered ankle inversion/eversion (Panzenbeck and Klute, 2012). Such a device has a potential of increasing balance (Panzenbeck and Klute, 2012), and as a result, providing the benefit of reducing stress on the residual limb (Portnoy et al., 2012). However, one study with a commercially available split-toe device did not find a benefit in dealing with a medial-lateral disturbance (Segal and Klute, 2014). The benefit of inversion actuation may be clearly revealed by exploring a wider range of inversion/eversion torques.

A simple way to apply ankle inversion torque is to create a stiffness in the ankle-inversion degree of freedom. This stiffness-based actuation is straightforward to find and implement in commercial device (Lindhe, 2014). By testing a wide range of stiffness including stabilizing and destabilizing actuation (Kim and Collins, 2015b), the effect of ankle inversion torque on balance would be clearly understood. However, different prosthetic stiffnesses can alter average applied torque, which may affect overall behavior, as in the case of adding work or torque to a walking device (Caputo and Collins, 2014a; Jackson and Collins, 2015). For this reason, it would be desirable to maintain a constant average torque over the different stiffnesses by modulating the ankle inversion torque proportional to the ankle angle deviation from a nominal value.

With a robotic ankle-foot prosthesis emulator, various stiffnesses can be implemented with zero average ankle inversion/eversion torque. We recently developed a two degree-of-freedom prosthesis, which can track a wide range of ankle inversion/eversion torques while maintaining consistent ankle plantarflexion behavior (Collins et al., 2015a). This type of device allowed us to implement various control ideas and determine the effect of ankle control on various balance-related effort (Kim and Collins, 2015b). The versatility of this multi-actuated prosthesis emulator would also enable tightly-controlled tests of the effects of inversion/eversion stiffness on balance during walking.

Balance-related effort can be measured by metabolic rate, step width variability, average step width, variability of the center of pressure beneath the intact foot, intact-limb muscle activity variation, and user preference. Active balance effort often requires extra energy consumption (Kim and Collins, 2015b; O'Connor et al., 2012). Additional energy consumption seems to relate to increases in active foot placement (O'Connor et al., 2012; Kim and Collins, 2015b) and step width (Donelan et al., 2001). Subjects also seem to increase control effort in the intact limb to maintain balance (Hof et al., 2005; Kim and Collins, 2015b). This intact limb control effort may also be revealed by variability in muscle activity (McGowan et al., 2009). Overall sense of balance and balance-related effort can also affect user preference scoring of a device (Kim and Collins, 2015b; Caputo et al., 2015a).

In this study, we hypothesized that balance-related effort would decrease when using a prosthesis controlled to provide a restoring torque proportional to deviation from a nominal inversion angle trajectory. In contrast, we hypothesized that a controller with the opposite behavior would cause an increase in balance-related effort. We also hypothesized that the effort could be revealed by at least one of six measures: metabolic rate, step width variability, average step width, center of pressure variability in intact limb, muscle activation variability in intact limb, and user preference. We expected that the results of this study would help guide the design of a prostheses that improve balance during walking.

6.2 Methods

We designed an inversion/eversion stiffness controller and implemented it on an ankle-foot prosthesis emulator system. We performed experiments in which unilateral transtibial amputees experienced different stiffness values across walking conditions. Metabolics, gait mechanics and satisfaction data were collected and tested for an effect of inversion stiffness.

6.2.1 Experimental hardware

We used a two degree of freedom robotic ankle-foot prosthesis emulator in our experiments. The lightweight prosthesis end-effector (0.72 kg) has two independent toes, enabling measurement of ankle inversion/eversion angle and application of

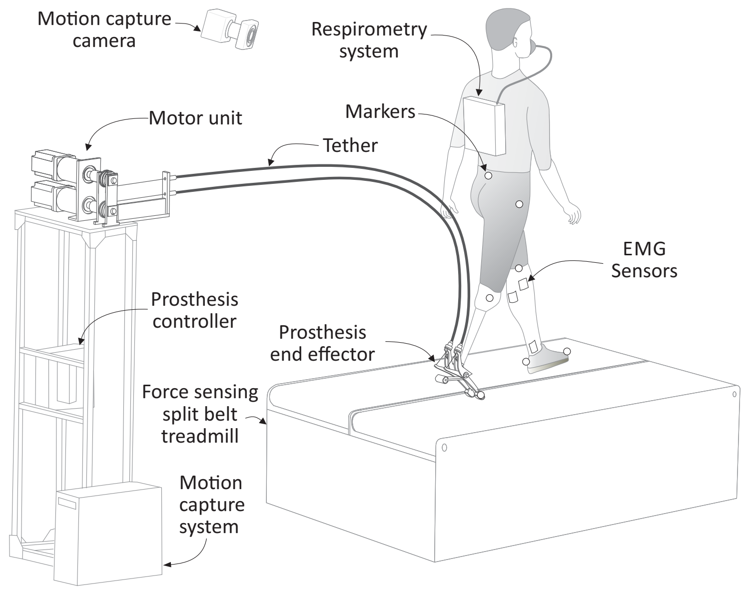

Figure 6.1: Overall experimental method. The subject wore the 2DoF prosthesis end effector while walking on an instrumented treadmill. The prosthesis was composed of two independently actuated toe end effectors, tethers, motors, and prosthesis controller. We varied ankle inversion/eversion stiffness using the system. We measured electromyography (EMG), metabolic rate using respirometry system, and the foot placement using a motion capture system and the instrumented treadmill.

ankle inversion/eversion torque, independent from plantarflexion behavior. The tethered emulator was powered by two off-board servomotors, which were connected to the rear of each toe via Bowden cables (Fig. 6.1). Control was performed via an off-board real-time control system (dSPACE Inc., Wixom, MI). This emulator is described in detail in (Collins et al., 2015a). The emulator provided appropriate features to test our hypothesis, including high torque, fast response to torque commands, and low sensitivity to disturbances.

Body position was detected using an instrumented treadmill, motion capture system, and sensors on the emulator. A split-belt treadmill with integrated force plates was used to determine both the force and moment applied to the treadmill in all three directions (Bertec Co. Columbus, OH, USA). Data was sampled at 1000 Hz and low-pass filtered at 100 Hz. Subjects were outfitted with 19 motion capture markers (sacrum and left and right sides of asis, greater trochanter, medial and lateral epicondyles of the knee, medial and lateral maleoli of the ankle, calcaneous, and first and fifth metatarsals of the foot). Markers were tracked by seven motion capture cameras operating at 100 Hz (Vicon, Oxford, UK).

Muscle activity of the intact lower limb was measured using electromyography sensors (Delsys, Natick, MA). Electrodes were placed on the medial and lateral aspects of the soleus, the medial and lateral gastrocnemius, the tibialis anterior and the peroneus longus.

Stance phase was detected using the instrumented treadmill, motion capture system, and the prosthesis emulator. For data analysis, foot contact was identified using the vertical component of force from the treadmill with a threshold of ten percent of the body weight. Erroneous data points from crossover steps were corrected through inspection of heel and toe marker positions. For on-line stance detection, we used a strain gauge on the heel of the prosthesis end-effector to detect heel strike. Once heel contact was signaled, toe contact was detected using an on-board torque sensor with a threshold of 2 Nm. The end of stance phase was detected using the value from the force plate with a threshold of 10 N. Thresholds were further tuned based on each subjects weight and gait pattern. Errors due to crossover steps and noise were eliminated using secondary thresholds on emulator torque and timing information.

6.2.2 Prosthesis Control

We applied an ankle inversion/eversion torque proportional to the ankle inversion/eversion angle deviation from a nominal value (eq. (1), Fig. 6.2). We varied the proportional gain (stiffness) with an intention of stabilizing or destabilizing the user's gait. The torque was applied during the stance phase.

$$\tau_{inv} = K \cdot (\theta(i) \; - \; \theta_{nom}(i)) \tag{6.1}$$

Where τ_{inv} is inversion torque, K is stiffness, θ is ankle inversion/eversion angle,

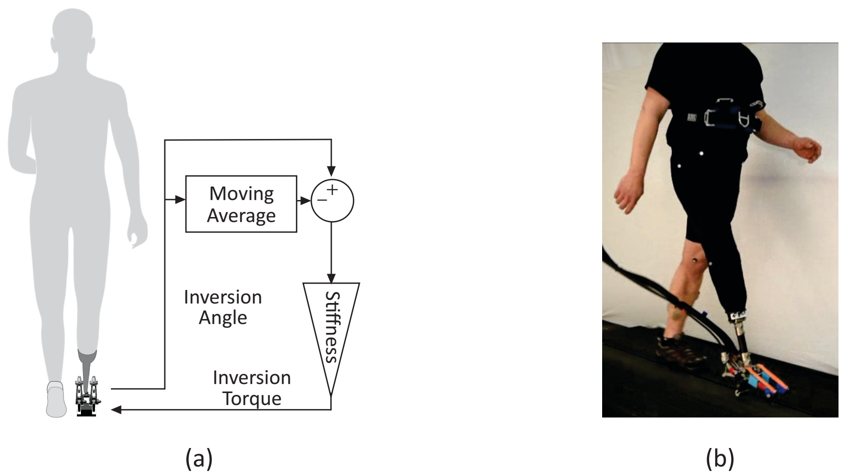

Figure 6.2: (a) Ankle inversion torque was applied by multiplying stiffness to measured inversion angle deviation from nominal values. (b) We conducted experiment with individuals with below knee amputation using a 2 DoF prosthesis.

and θ_{nom} is the nominal angle. i is an index during stance phase.

The nominal ankle inversion/eversion angle was learned on-line during a stance phase. We averaged the inversion angle at each instance of a stance phase by adding the current inversion angle to the previously learned inversion angle (eq. (2)).

$$\theta_{nom}(i, n+1) = \alpha \cdot \theta(i, n) \; - \; (1-\alpha) \cdot \theta_{nom}(i, n) \tag{6.2}$$

Where θ_{nom} is the nominal inversion angle, θ is the measured inversion angle, α is a learning gain, i is the time stamp during a control cycle within current stance phase, n is the current stance phase, and $n+1$ is the stance phase on the next walking step.

Ankle plantarflexion torque was applied by providing ankle torque as a function of ankle plantarflexion angle. We approximated the human ankle angle-torque curve as linear piecewise curves, composed of dorsiflexion and plantarflexion sections (Caputo and Collins, 2014b). The impedance of these curves was varied depending on user preferred stiffness. The plantarflexion part was modified to supply the user-preferred work.

6.2.3 Experimental Protocol

This study was conducted using five subjects with unilateral transtibial amputation (N = 5, all male, all traumatic, all K3 ambulators, 4 left-side amputation, age = 47.8 ± 12.8 years, body mass = 83.9 ± 13.9 kg, height = 1.78 ± 0.041 m, mean ± s.d.). This study was conducted in accordance with a protocol approved by the Carnegie Mellon University Institutional Review Board.

Seven conditions were presented per collection: five stiffness conditions, one prescribed prosthesis condition, and one quiet standing condition. The stiffness conditions were composed of Stabilizing High stiffness, Stabilizing Low stiffness, Neutral, Destabilizing Low stiffness and Destabilizing High stiffness with the corresponding numerical values of 5, 2.5, 0, −2.5, −5 N·m·deg^{-1}, respectively. We expected that the conditions with positive stiffness would reduce balance-related effort, while the conditions with negative stiffness would increase balance-related effort, in comparison to the zero stiffness condition. The prescribed prosthesis condition offered baseline walking data. The quiet standing condition provided baseline metabolic data.

Subjects walked for six minutes at 1.25 m/s, experiencing five conditions on this device and one trial on their own device. We reduced the duration to a minimum of four minutes and the speed to a minimum of 0.75 m/s for two subjects who had lower activity levels. Between trials, subjects had four to ten minutes rest, depending on their activity level. The five conditions were randomized. We also randomly placed the prescribed prosthesis condition either at the beginning or at the end of trials. Before starting the five stiffness conditions, subjects participated in a three-minute quiet standing trial while wearing the emulator. Subjects experienced these conditions for one training day and one data collection day. We present data of the last two minutes of steady state walking during the data collection day.

In total, subjects participated in at least four days of study: pylon fitting and acclimation, user-specific parameter optimization, training, and data collection. On the first day of the study, a Certified Prosthetist fit a pylon to connect from the subjects typical socket to the ankle-foot emulator. During the second day, parameter fitting was conducted based on user preference using an established process of variable triangulation (Caputo et al., 2015b). We optimized the plantar-dorsal stiffness, neutral angle and push-off work parameters. Subjects that were new to the device were also given an additional day of acclimation in order to become more comfortable on the emulator. On training and data collection days, subjects went through the seven conditions with randomization as described above.

6.2.4 Measures of balance-related effort

We measured metabolic energy consumption, average step width, step width variability, center of pressure variability during intact-limb stance, and muscle activity in intact limb.

Metabolic energy consumption was sensed and recorded using a wireless metabolic unit (Oxycon Mobile, CareFusion, San Diego, CA, USA; Fig. 6.1). Steady state oxygen consumption was determined by the plateau in oxygen consumption. The net average metabolic rate was calculated using a standard equation (Brockway, 1987) and subtracting the metabolic rate found during the quiet standing period.

Step width was determined using a combination of the trajectories of the motion capture markers and the center of pressure information from the instrumented treadmill. Step width was defined as the mediolateral distance between left and right foot placements during mid-stance. The marker-based foot location was calculated as the average of the toe and heel markers at mid-stance. The center of pressure-based foot location was also calculated at mid-stance. Mid-stance was specified as the time when the marker on the sacrum was directly above the marker on the stance heel in the sagittal plane. Marker and treadmill data were low-pass filtered at 20 Hz before calculating step width for each step. The average step width and step width variability were defined as the mean and standard deviation, respectively, of step width across all steps in the collection window.

Center of pressure variability was determined throughout the stance phase using treadmill data using the method of Kim and Collins (2015b). At each instant of stance, the mediolateral location of the center of pressure was found and the average center of pressure location was subtracted from this value. This resulted in a trajectory for each step that was normalized in time. Then, the standard deviation of the center of pressure location at each fraction of the stance was calculated and averaged across all moments in the stance.

Each subject was also asked to rate the trial immediately after the walking period on a scale of -10 to 10. Subjects were instructed to calibrate their responses assuming that 0 represented walking with their prescribed device, -10 representing a condition that is impossible to walk in, and 10 representing an optimal, effortless walking condition.

EMG activity variability was measured by calculating standard deviation of EMG activity at each instant of time during stance period. EMG signals were high pass filtered, rectified, and then low pass filtered with a frequency of 10 Hz and 6Hz, respectively (Ferris et al., 2006). We normalized EMG activity in time during stance phase and we subtracted the average EMG activity for each step. Standard deviation of the EMG activity across steps was calculated and the values were averaged.

6.2.5 Statistical analysis

We tested for an effect of stiffness on biomechanic outcomes with repeated-measures ANOVA with a significance level of $\alpha = 0.05$. For significant outcomes, we performed paired t-tests among each controller condition with a significance level of 0.05. We also performed a post-hoc power analysis with a power level 0.7 to estimate the correct sample size.

6.3 Results

The controller provided restoring or diverging torque at each step while keeping the average torque at zero. Foot placement variability, average step width, and intact limb center of pressure variability were reduced for the stabilizing stiffness conditions compared to the destabilizing stiffness conditions. Users preferred the stabilizing stiffness conditions over destabilizing stiffness conditions. Metabolic energy consumption and evertor muscle variability in intact limb side did not show any trend.

The prosthesis applied ankle inversion/eversion torque proportional to the deviation of ankle inversion/eversion angle from the nominal trajectory (Fig. 6.2(a)), while maintaining average torque as zero across entire collection (Fig. 6.2(b)). The root mean square error of the torque tracking was within 3 Nm. Zero average torque was obtained as a result of applying torque deviation from the learned neutral angle, which varied slightly in time and differed slightly across conditions (Fig. 6.2(c))).

Variability in step width significantly decreased with increasing stiffness (ANOVA, $p = 0.0003$; Fig. 6.4(a)). Step width variabilities were reduced 28% and 22% for the Stabilizing high stiffness condition and Stabilizing low stiffness condition compared to Destabilizing high stiffness condition ($p = 0.008$ and 0.017, respectively). Post-hoc power analysis showed that five subjects were enough for this paired t-test.

Average step width was reduced for stabilizing stiffness conditions (ANOVA, $p = 0.004$; Fig. 6.4(b)). High restoring stiffness reduced the average step width by 12% compared to the Destabilizing high stiffness conditions ($p = 0.006$). According to post-hoc power analysis, four subjects were enough for this paired t-test.

Center of pressure variability was reduced for stabilizing stiffness conditions compared to destabilizing conditions (ANOVA, $p = 0.005$; Fig. 6.4(c)). The

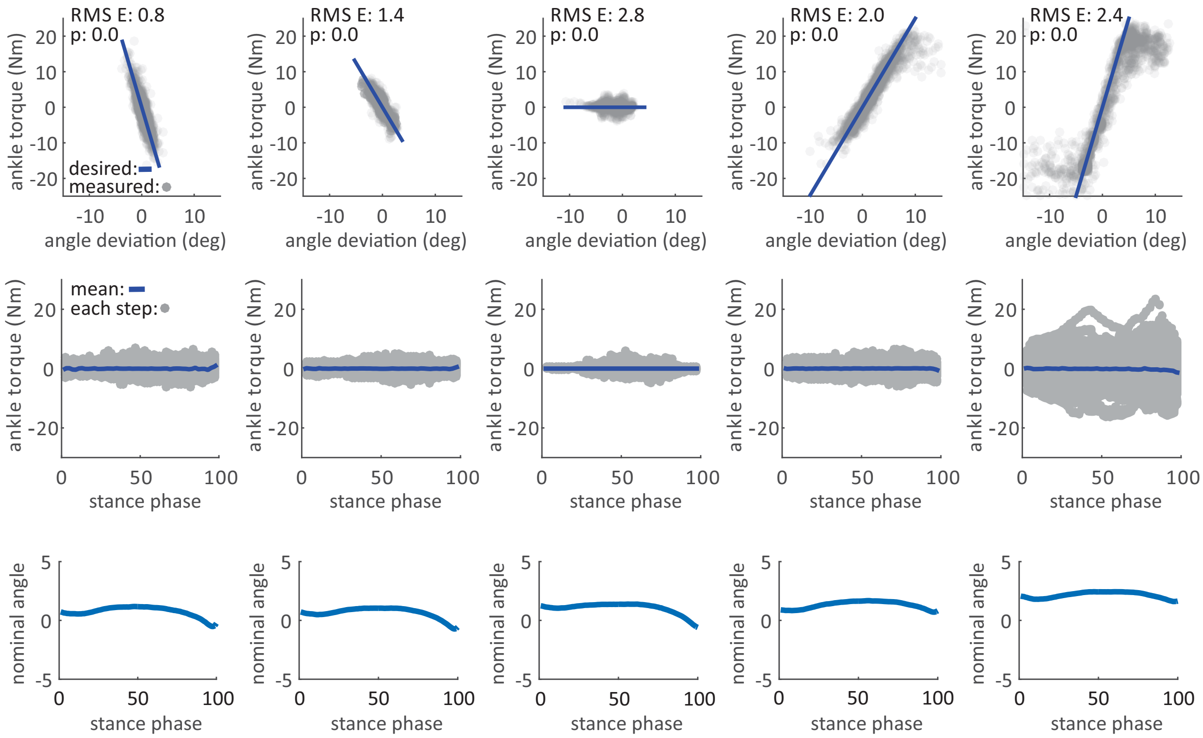

Figure 6.3: Ankle inversion/eversion torque profile as a function of ankle inversion/eversion angle (a) and as a function of time (b), and nominal ankle inversion/eversion angle during stance phase of a representative subject (c). RMS E stands for root mean squared error in Nm. p is the p-value for the F statistics, examining zero coefficient. This controller provided ankle torque proportional to ankle angle while maintaining an average zero torque. The subject showed a greater inversion angle path for Detabilizing high stiffness condition.

Stabilizing high stiffness and Stabilizing low stiffness conditions reduced the variability by 29% and 20% compared to the Destabilizing high stiffness condition, respectively (p = 0.025 and p = 0.024). According to a post-hoc power analysis, five subjects were enough for this t-test.

Subjects preferred stabilizing stiffness conditions compared to destabilizing stiffness conditions (ANOVA, p = 0.0003; Fig. 6.4(d)). Subjects preferred the Stabilizing high stiffness, Stabilizing low stiffness, and Neutral stiffness conditions 81%, 86%, and 90% more than the Destabilizing high stiffness condition (p = 0.025, 0.032, and 0.012, respectively).

Metabolic rate and intact limb inverter EMG data did not show a trend (ANOVA,

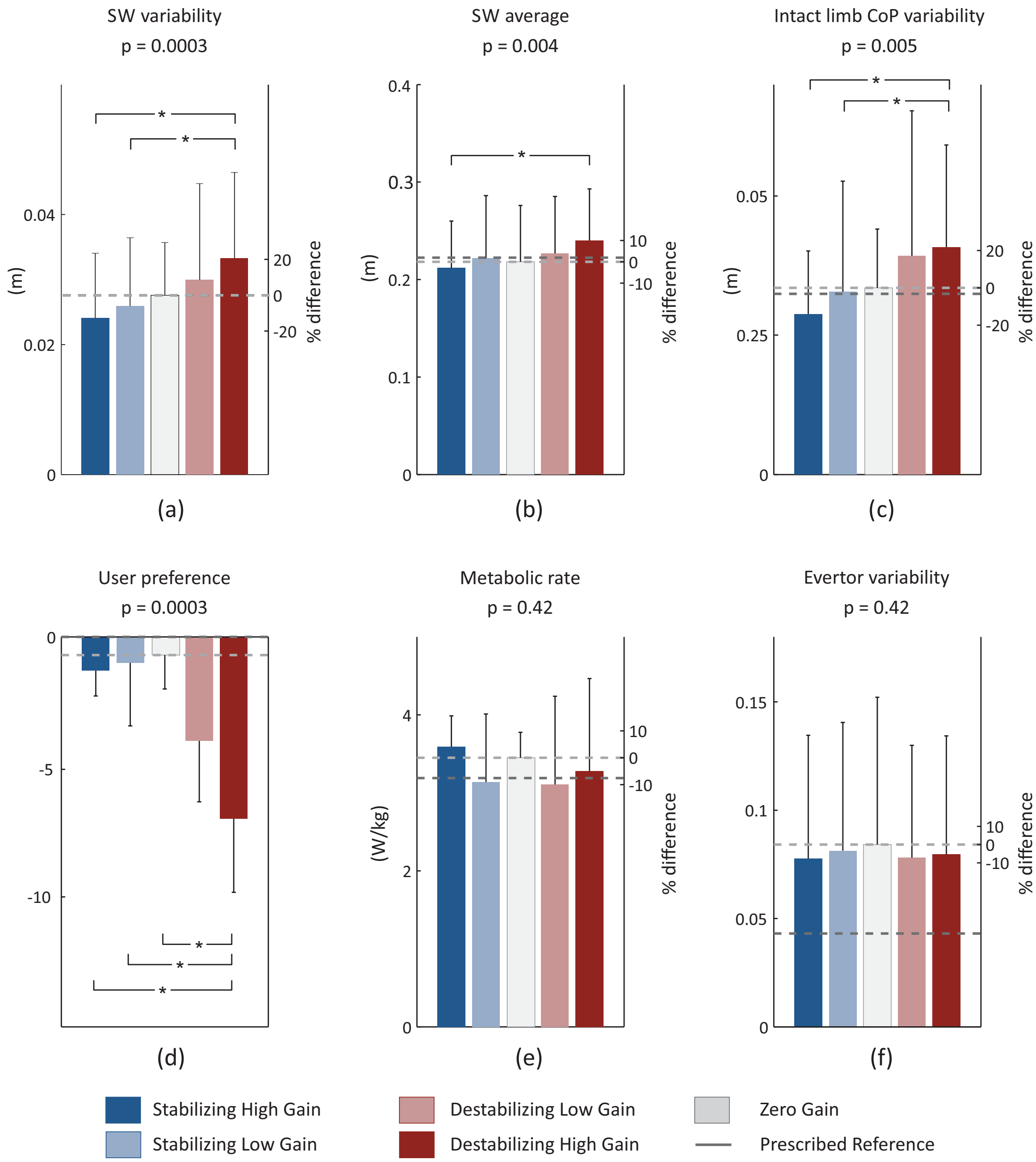

Figure 6.4: Balance-related measures: step width variability (a), average step width (b), intact limb center of pressure variability during stance phase (c), user preference (d), metabolic rate (e), and everter (peroneus longus) muscle variability (f). p is p-value for ANOVA analysis. Step width variability, average step width, intact limb center of pressure variability during stance and user preference were significantly affected by stiffness. Asterisks indicate statistical significance for pair-wise comparisons of conditions.

$p = 0.42$ and 0.42, respectively; Fig. 6.4(e,f)).

The Destabilizing high stiffness condition increased step width variability and center of pressure variability by 21% and 26% compared to walking with the prescribed prosthesis ($p = 0.036$ and $p = 0.023$, respectively). Subjects also disliked the controller compared to their prescribed prostheses ($p = 0.005$). For this t-test, six and five subjects were necessary for each measure.

6.4 Discussion

We investigated the effect of ankle inversion/eversion stiffness on balance-related effort. The controller applied restoring or diverging inversion/eversion torque at each step without affecting average inversion/eversion torque or plantarflexion torque. In response to stabilizing controllers, subjects reduced indicators of active foot placement, average step width, and intact limb center of pressure control effort during the stance phase. Subjects also preferred these controllers over the destabilizing ones. These results support the hypothesis that positive inversion/eversion stiffness reduces balance-related effort on level ground.

The stabilizing controllers seem to reduce balance-related hip actuation effort, shown by the reduction in average step width and step width variability (Fig. 6.4). At each stance phase, the stabilizing stiffness controller applied restoring torque, proportional to deviation from nominal angle (Fig. 6.3). Using this assistance, subjects might be able to reduce overall hip actuation effort to maintain balance by decreasing average step width (Curtze et al., 2011; Gates et al., 2013b; Hof et al., 2005). This decrease in step width might be correlated to decrease in ankle inversion/eversion angle (Fig. 6.3). This stabilizing stiffness controller also might allow more natural swing motion, shown by reduction in step width variability (Collins and Kuo, 2013), opposite to the result of destabilizing stiffness condition. These efforts seemed to be even more than 10 % lower than the condition of their prescribed prosthesis ($p = 0.4$ for both measures).

This stabilizing stiffness also might help to reduce balance-related ankle control effort. After foot placement, balance can be further adjusted by through movement of the stance-side ankle (Hof et al., 2007; Kim and Collins, 2015b). This active ankle control strategy seems to be more heavily used for the diverging stiffness condition compared to the stabilizing stiffness condition, as indicated by higher center of pressure variability in intact limb (Fig. 6.4). This result further supports our hypothesis that ankle inversion/eversion control is a method to maintain balance, and providing stabilizing stiffness may help to reduce balance-related effort.

In addition to the benefits of the reduction in efforts, users preferred the stabilizing controller to the destabilizing controller. User preference has been shown to be an important measure of prosthesis performance (Caputo et al., 2015a), and stabilizing controllers seemed to positively impact preference (Kim and Collins, 2015b). In this study, we also found that subjects strongly disliked the destabilizing stiffness condition compared to other controller conditions. This result suggests that this kind of device may satisfy the desire for a balance-assistive device (Legro et al., 1999).

Changes in average ankle torque may cause different balance-related efforts, so the torque was controlled across trials. Alteration of average controller behavior can affect effort during walking (Caputo and Collins, 2014b). In our experiment, average torque, however, was maintained as zero, while torque at each stance point was varied proportional to the deviation of the ankle angle from the nominal value (Fig. 6.3). This result clearly indicates that the difference in balance-related effort was caused by the restoring behavior of the virtual stiffness, rather than average torques as a result of nominal device alignment.

While previous experimentation showed a reduction in metabolic rate for the stabilizing controller conditions (Kim and Collins, 2015b), this study did not show a trend in metabolic rate. There are two possible explanations for this discrepancy –

muscle activation strategy and upper body motion. Muscle activation strategy has been shown to vary greatly among different age groups (Schmitz et al., 2009) and type of amputation (Wentink et al., 2013). Amputation is also associated with high inter-subject variability in muscle activity (Wentink et al., 2013; Alcaide-Aguirre et al., 2013), which may partially explain a lack of trend in muscle activation in our study (Fig. 6.4). This difference in muscle activity may also affect metabolic rate (Bisi et al., 2011; Blake and Wakeling, 2013) and explain the difference compared to the previous study (Kim and Collins, 2015b). More training may normalize this difference (Alcaide-Aguirre et al., 2013) by reducing unnecessary co-contraction (Moore and Marteniuk, 1986) and therefore, reducing metabolic energy.

Metabolic rate can also be affected by upper body motion, which may be used to maintain balance. Upper body motion, such as arm swing or trunk motion, has been observed as a balance resource during walking among individuals with below knee amputation (Curtze et al., 2011; Lamoth et al., 2010; Ramstrand and Nilsson, 2009), as opposed to foot placement control on uneven ground (Collins and Kuo, 2013). Subjects may have used these other upper body strategies to maintain balance, which may partially explain the lack of trend in metabolic rate.

While the highest stabilizing control condition brought about a great reduction in effort, a full exploration of controller conditions is necessary. Unexpectedly, the largest reduction in balance-related effort occurred for the condition with the highest stabilizing stiffness, which was slightly higher than the stiffness found in the pilot study. To understand this result, we conducted a simulation study using a previously developed model (Kim and Collins, 2013). More specifically, we examined the maximum tolerable ground height for different models with various ankle inversion/eversion stiffnesses under foot placement control. The simulation results did not indicate the most stable condition to be that of the high stabilizing stiffness. Perhaps this experimental outcome was caused by training effect, similar to the previous study (Kim and Collins, 2015b). By providing a wider range of conditions, we may have improved our understanding of the influence of stiffness on balance-related effort on even ground. However, this experimental design might be complicated because of higher inter-subject variability, associated with amputation (Huang and Ferris, 2012; Alcaide-Aguirre et al., 2013). A possible solution could be a user-specific online optimization method (Caputo et al., 2015b).

In summary, this study suggests that moderate positive stiffness is beneficial for amputees, even on level ground. By applying this finding to a commercial prosthesis, subjects may increase balance, and therefore, increase social activity (Lin et al., 2014). Future research should conduct experiments on uneven ground as this high stiffness might be unfavorable for uneven ground.

Chapter 7

The ankle inversion/eversion torque may reduce balance-related effort of unilateral, transtibial amputees

In this study, I examined the effect of active, as opposite to passive, ankle inversion control on balance-related effort. I conducted an experiment on individuals with below knee amputation (N = 4) to examine the effect of a step-to-step ankle inversion/eversion torque modulation on balance-related effort. The controller provided step-to-step restoring torque proportional to a lateral acceleration deviation from the nominal value, while average torque was kept constant throughout the trial.

I found that, compared to the destablizing controller, the stablizing controller can reduce balance-related effort, as shown by lowered metabolic rate. This reducetion was also shown in three subjects when compared to the no controller condition.

This reduction in overall energy consumption was shown from the first day; these first day benefits suggest that an ankle inversion/eversion torque controller might be transparent. Experimenting with more subjects may more clearly reveal such gain.

The contents of this chapter will appear in:

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A preliminary version of this work was presented at:

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Abstract

Below-knee amputation is associated with greater levels of balance-related effort. This effort can be reduced by controlling push-off work at each step. The push-off work control affects in fore-aft and medio-lateral motions. The lateral motion can be more directly controlled by ankle inversion/eversion torque modulation at each step. This control idea initially confirmed by a simulation study using a three-dimensional limit cycle walking model. Motivated by simulation result, I conducted group and single-case experiments with four amputees. During a typical group experimental session, we tested five controller conditions: two stabilizing controllers, a zero gain controller, and two destabilizing controllers. Stabilizing controllers applied ankle inversion torque during stance phase when higher lateral acceleration toward the intact limb side was detected at the intact limb toe-off instant. We hypothesized that this controller would reduce balance-related effort. Destabilizing controllers applied an increase in eversion torque in response to acceleration in the inward direction, thus increasing the effort needed to balance. During a single-case experimental session, we tested a zero gain and a stabilizing controller. We hypothesized that a single-case experiment would reveal the effect of the controller on some subjects. In addition, we hypothesized that balance-related effort would be revealed by one of the following measures: metabolic rate, intact limb center of pressure variability, step width variability, average step width, and user preference. We found that during group experimental sessions, the stabilizing controller reduced metabolic rate compared to the destabilizing controller from the first day. The subjects also preferred the stabilizing controller over the destabilizing controller. For the single-case experiment, three subjects reduced metabolic rate by 6%, 11%, and 14% compared to the no controller condition after forced exploration. These results show that this step-to-step ankle inversion/eversion controller has the potential to reduce balance-related effort.

7.1 Introduction

Individuals with below knee amputation exert more effort during walking than their able-bodied counterparts (Waters and Mulroy, 1999). This increased effort may partially be due to increased balance-related effort, especially while walking on challenging terrain (Paysant et al., 2006). Recently, we found that balance-related effort can be reduced by modulating ankle push-off work at each step with a robotic ankle-foot prosthesis, an experiment which was motivated by our previous simulation study (Kim and Collins, 2015b). These studies also suggested a possible balance-restoring resource, ankle inversion/eversion control.

Active ankle inversion/eversion control may help stabilize lateral motion during walking. Human walking seems to be the least stable in the medio-lateral direction (Kuo, 1999), which is also the direction that requires more effort to maintain balance for below knee amputees (Segal and Klute, 2014). This effort was reduced for some individuals with a step-to-step push-off work controller, which resulted in reduced active ankle inversion control effort (Kim and Collins, 2015a). The possibility of using active ankle inversion control as a balance restoring method was also shown through simulation study (Kim and Collins, 2015b). In the study, ankle inversion resistance modulation improved balance, even though the performance was inferior to push-off work modulation. While both the push-off work controller and ankle inversion resistance controller affect lateral motion, the indirect effect of the control action on restoring balance in the medio-lateral direction may require an extended learning process (Kim and Collins, 2015a) or may decrease disturbance tolerance (Kim and Collins, 2015b). Perhaps by providing ankle inversion/eversion controller with more straightforward control action from a lateral disturbance, amputees may perceive more direct benefit in balance.

Once-per-step ankle inversion constant torque modulation

One possible method of direct control could be modulating ankle inversion/eversion torque at each step. This control strategy could directly facilitate redirection of the center of mass velocity when it deviates from the nominal value at intact limb toe-off instant by applying a force in the opposite direction. This control method seems to be used by humans during walking to recover from external disturbances or slightly imprecise foot placement (Hof et al., 2010).

The feasibility of step-to-step ankle inversion torque modulation can be evaluated by a simulation study before conducting a full human-subject experiment. The simulation study provides a platform to rigorously compare the effect of the proposed controller with other lateral stabilization methods (Kim and Collins, 2013). If the performance of the inversion/eversion torque controller is comparable to other stabilization methods, such as foot placement control and ankle push-off work modulation, then active inversion/eversion constant torque modulation in an ankle-foot prosthesis could be a viable solution to improve balance.

However, due to assumptions and simplifications used in the simulation, it is also important to run an experimental study exploring the effect of ankle inversion/eversion torque control on stability, using group and single-case experimental protocols. A typical group experiment could be used to show the effect of an easily understandable controller on balance (Kim et al., 2015). In the previous study, however, this design failed to show the efficacy of one of the important measures: metabolic rate. This result might be caused by high inter-subject variability (Quesada et al., 2015), which can mask the effect of the controller when averaging group response. By investigating each subject's response as well as group response using single-case experimental design, the efficacy of the controller might be more clearly revealed (Dermer and Hoch, 2012). Additionally, a forced exploration period can help each individual to better learn how to use the controller (Selinger et al., 2015) to reduce balance-related effort (Kim and Collins, 2015a).

When designing such experiments, the level of each control action needs to be determined. The degree of the action can be manipulated by selecting gains for each controller. The gains could be decided based on pilot study results (Kim and Collins, 2015b). This method may suggest using multiple gains to account for inter-subject variability, which is implementable in a group experiment. For a single case experiment with forced exploration, using several gains is not practical because of prolonged length of trial. Instead, the gain could be further optimized for each subject based on previous trial results by examining measures of balance-related effort (Kim and Collins, 2015a). We hypothesized that the gain would be selected based on two indicators in balance-related effort to reduce the effect of noise. We also hypothesized that the more important measure would capture gross effort since each subject may use different balance strategy (Kim and Collins, 2015b,a; Curtze et al., 2011). In addition, slightly higher gain might provide more benefit, considering the continuous learning ability of a subject (Kim and Collins, 2015b).

Analysis of the effect of controller on balance-related effort

The effect of the controller might be shown through indicators of balance-related effort: metabolic rate, step width variability, average step width, intact limb center of pressure variability, user preference in balance and comfort (Kim and Collins, 2015b). Metabolic rate may reveal altered overall muscle activity to maintain balance. Step width variability may be associated with foot placement effort. Intact limb control effort may show active ankle inversion/eversion control effort. User preference may be associated with the user's perceived benefit from the controller . By separating user preference in balance and comfort, we may be able to split other effects than step-to-step controller, such as push-off work (Quesada et al., 2015).

Using these measures, each subject's response of the step-to-step inversion torque controller can be analyzed using both single-case and group experimental results. Inspecting individual data from single-case experiment may reveal the specific effect of the controller for some subjects. The predetermined order of a single-case experiment, however, may cause an ordering effect, which may decrease the reliability and significance of this controller. This ordering effect could be reduced by averaging same conditions with a lag (Gentile and Klein, 1972). When significant ordering effect is observed, the reliability of an intervention needs to be more rigorously evaluated (Hartmann, 1974). Although a solution is statistical analysis of single-case experiments with a random order (Edgington, 1967, 1969, 1972), the randomization of conditions can prolong trials. Perhaps by running a statistical analysis across several days, the reliability of the controller might be practically examined, even though the final day trial could be conducted in a predetermined order.

Statistical analysis of group experiment may show the external validity of this controller. Single-case experiments may exhibit the efficacy of the controller for the individuals who received the benefit significantly and reliably. This result, however, may not be applicable to other subjects. The external validity can be shown by conducting statistical analysis on group responses. For this analysis, we may be able to use fully randomized group experiment results if subjects can utilize the controller during the period. This might be possible because the step-to-step ankle inversion controller directly impacts side-to-side motion (Kim et al., 2015; Kim and Collins, 2015a).

Study aims and hypothesis

In this study, we explored the effect of step-to-step ankle inversion/eversion torque control on balance through simulation and human-subject experimentation. We hypothesized that active ankle inversion/eversion control at each step would be relatively effective at restoring balance compared to other stabilization methods. We also hypothesized that stabilizing step-to-step ankle inversion/eversion torque modulation in a robotic ankle-foot prosthesis would reduce balance-related effort in some individuals with below knee amputation, while a destabilizing controller would increase effort. We expect the results of this study to offer insights into potential control methods for reducing balance-related effort for individuals with below knee amputation.

7.2 Methods

We initially examined the effects of step-to-step ankle inversion/eversion controllers on balance using a simulation study of a three-dimensional limit cycle walking model. Then, we conducted experiments with four below-knee amputee subjects and measured indicators of balance-related effort including metabolic rate, step width variability, average step width, center of pressure variability in the intact limb, and user preference. The experiments were designed to investigate group response and each subject response.

7.2.1 Simulation methods

We used a three-dimensional limit cycle walking model with hip and ankle joints to compare the effect of different modes of actuation at each of the joints on balance recovery (Kim and Collins, 2013). The ankle inversion/eversion joint was actuated to apply a constant torque during the stance, while the foot was flat on the ground (Fig. 7.1(a)). The ankle plantarflexion joint was controlled to provide push-off work. The hip abduction joint was modulated quasi-statistically to change step-width, and the hip flexion joint was actuated to track desired step length.

To stabilize the model during walking, we designed discrete linear feedback

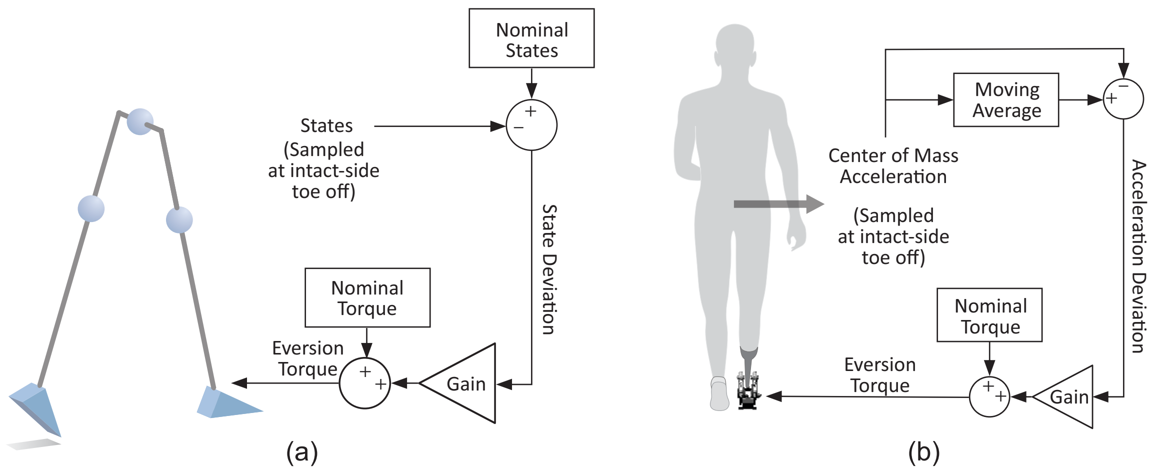

Figure 7.1: (a) controller design simulation: ankle inversion/eversion torque was modulated once per step, by multiplying state deviation with a gain. (b) controller design for experiment: the magnitude of ankle inversion/eversion torque was modulated, proportional to the deviation of center of mass acceleration to nominal acceleration.

controllers, which altered actuation parameters once per step. We controlled parameters with high control authority in the medio-lateral direction (the magnitude of inversion/eversion torque (Fig. 7.1(a)), the magnitude of push-off work, step-width, and the combination of step-width and inversion/eversion torque). The gain of the controller was optimized using the LQR algorithm. The performance of each controller was quantified by evaluating two measures while the model walked one hundred steps without falling: 1) maximum tolerable random disturbance in lateral impulse and 2) maximum tolerable random disturbance in ground height.

7.2.2 Experimental methods

We conducted an experiment with four participants with below-knee amputations with group experimental protocol and single-case experimental protocol. Step-to-step ankle inversion/eversion torque was provided using a robotic ankle-foot prosthesis. Subjects walked on an instrumented treadmill while wearing metabolic rate measurement equipment, markers, and electromyography (EMG) sensors similar to previous study (Kim et al., 2015). We measured balance-related effort.

The data were processed to uncover the effect of controller for each subject and also for other amputees, outside of this experimental pool. We used both visual inspection and statistical analysis to evaluate the results.

Prosthesis control

Emulator: We used a two-degree of freedom ankle foot prosthesis to test the performance of our controllers on below-knee amputees. This testbed (described in detail in (Collins et al., 2015a)) was powered by two off-board motors, which enabled the prosthesis to weigh only 0.72 kg. The device had two independently actuated toes, which allow for a combined plantar flexion torque of up to 180 Nm and an ankle inversion/eversion torque of ± 30 Nm with a peak power of 3 KW. It also had a torque bandwidth higher than 20 Hz. Leveraging these characteristics, we were able to test the controllers robustly over a wide range of operating parameters.

Controller design: The controller is composed of a high level step-to-step controller and a low level continuous controller. The high level controller decided the desired inversion/eversion constant torque at each step and the low level controller controlled the actual torque delivered by each toe.

High level step-to-step control: The step-to-step controller chose an ankle inversion/eversion torque on each step as a linear function of the estimated medio-lateral acceleration (Fig. 7.1(b)). More specifically, the medio-lateral acceleration was determined using lateral force information from the force plate treadmill (Bertec Co. Columbus, OH, USA). The value was obtained by dividing the sum of right and left lateral forces by the subject's mass. The control decision occurred at the instant of toe-off of the contra-lateral limb.

Low level, continuous control: The low level controller calculated and controlled torque for each toe to meet the desired ankle inversion/eversion torque and plantarflexion torque. Desired inversion/eversion torque was commanded from high level controller and desired plantarflexion torque was estimated using a piece-wise, linear approximation of the human ankle work loop. Ankle angle was determined by motor encoders affixed to each toe of the prosthesis. Heel-strike of the prosthetic limb was determined using strain gauge mounted on the fiberglass heel and force data from the instrumented treadmill. Heel strike of the intact limb was determined using the instrumented treadmill force data. Ankle inversion/eversion and plantarflexion torque were determined using strain gauges affixed to the top and bottom of each toe.

Experimental protocol

Participants: Four subjects with below-knee amputation from traumatic incidents participated in this study (N = 4 all male, all traumatic, all K3 ambulators, 3 left-side amputation, age = 47.8 $\pm$ 14.3 years, body mass = 79.8 $\pm$ 12.7 kg, height = 1.71 $\pm$ 0.037 m, mean s.d.). All subjects had participated in previous studies with this device (Kim et al., 2015). Experimental protocol was approved by the Carnegie Mellon University Institutional Review Board and all experimentation were conducted under these guidelines.

Conditions: The subjects participated in three study sessions: an acclimation day, a group experiment period and gain selection day, and a single-case experiment period with forced exploration day. For the first and second day, subjects experienced five controller conditions: a Stabilizing high gain, Stabilizing low gain, Neutral, Destabilizing low gain, and Destabilizing high gain. For the single-subject experiment session, subjects were exposed to two controller conditions: a Stabilizing controller with predetermined gain and a Neutral controller. The default ankle inversion torque was found based on user feedback while sweeping torques ranging from -5 Nm to 5 Nm (in a randomized direction) (Caputo, 2015). Additionally, throughout sessions, each subject walked on his own prescribed device to obtain a baseline, randomly selected to occur either at the beginning or end of the session. Each session also included a three-minute quiet standing period while wearing the robotic device and a brief parameter check to re-familiarize the user and ensure all device parameters were comfortable.

The default walking period was five minutes with four minutes of rest between trials. For less-active subjects, the trial period was reduced to between four and five minutes and the rest time was increased to between four and ten minutes. The treadmill speed was set to 1.25 $m \cdot s^{-1}$, except for less active subjects, where it was determined that this speed was infeasible (here, the treadmill speed was adjusted between 0.7 $m \cdot s^{-1}$ and 1.25 $m \cdot s^{-1}$).

Day 1 - Acclimation period: On the first day of testing, seven conditions were presented: five controller conditions, one quiet standing condition, and one prescribed prosthesis condition. The five inversion/eversion controller conditions were divided into three blocks: one block with two stabilizing controllers, one block with one neutral controller, and one block with two destabilizing controllers. The order of the three controller blocks was randomized. For each stabilizing and destabilizing controller block, we presented the high and low gain controllers in the order of increasing gain. The gains were -30 $Nm \cdot (m \cdot s^{-2})^{-1}$ for Stabilizing high gain, -15 $Nm \cdot (m \cdot s^{-2})^{-1}$ for Stabilizing low gain, 0 $Nm \cdot (m \cdot s^{-2})^{-1}$ for Zero gain, 15 $Nm \cdot (m \cdot s^{-2})^{-1}$ for Destabilizing low gain, and 30 $Nm \cdot (m \cdot s^{-2})^{-1}$ for Destabilizing low gain.

Day 2 - Group experiment and gain selection period: The second day of testing consisted of the same walking conditions as the first session. However, five controller conditions were fully randomized.

Day 3 - Single subject experiment with forced exploration period: For the final day of testing, subjects experienced eight conditions: six controller conditions, one quiet standing period, and one condition with their prescribed prosthesis. For the six controller conditions, we presented Zero gain controller (A) and Stablizing controller with predetermined gain (B) repeatedly with the following order: A with forced exploration for four minutes, A with data collection for five minutes, B with forced exploration for four minutes, B with training for four minutes, B with data collection for five minutes, and A with data collection for five minutes. Between trials, we provided a rest between four to ten minutes. Data collection duration was adjusted to between four and five minutes to accommodate less active individuals.

On this collection day, the Stabilizing controller used a pre-determined gain based on the user's reaction from the previous trial. We compared the subject's response to the High stabilizing and the High destabilizing condition. Once we found a 10% reduction in metabolic rate and a 10% reduction in one of the other measures of balance-related effort (step-width variability, average step width, center of pressure variability in intact limb, or user preference), we considered that the gain used in that condition might be optimal. If the gain was the highest, then we used its value. If the gain was the lowest, then we used a 30% higher gain for the experiment with forced exploration, to account for a learning effect. If only the metabolic rate showed more than a 10% reduction for the stabilizing controller condition compared to destabilizing controller condition, then we also used a 30% higher gain than the lowest gain based on pilot study results. If a trend was not clear from the group experiment period, then we analyzed the acclimation period data. Similarly, if we observe more than 10% reduction in metabolic rate between the stabilizing controller condition and the destabilizing controller condition, then we applied 30% higher gain than the low gain. If the subject did not show any reduction in metabolic rate, then we used the lowest gain.

During the forced exploration portions, each subject was instructed to attempt different walking strategies for a total of three minutes. After walking normally for one minute, 12 different instructions were given, one every 15 seconds. The instructions included: “lean a little bit to your left,” “lean a little more to your right,” “sway more side to side,” “lean a little forward,” “lean a little backward,” “keep your body upright, use limited sway,” “take slightly wider steps,” “take slightly narrower steps,” “take slightly longer steps,” “take slightly shorter steps,” “swing your arms more than you typically would,” and “swing your arms less than you typically would.” For the less active amputees, the instructions “keep your body upright, use limited sway,” “take slightly shorter steps,” and “swing your arms less than you typically would” were omitted and instructions were given every 20 seconds, as opposed to 15 seconds. This was done in order to preserve the overall length of the trial and allow more time for these subjects to develop a gait based on the instructions. The second trial of the A block was a four- or five-minute period of uninstructed walking, followed by a five to ten minute rest (times depended on subject activity level).

Balance-related measures

In order to determine balance related effort, average step width, step width variability, intact-limb center of pressure variability, metabolic rate, and user preference were calculated (Kim and Collins, 2015b). Center of pressure variability was calculated using force and moment data from the force-plate instrumented split-belt treadmill. Data was sampled at 1000 Hz and low-pass filtered at 100 Hz. Step width variability and average step width were obtained using recorded data from 7 motion capture cameras (Vicon, Oxford, UK) using five motion capture markers (left heel, right heel, left toe, right toe and sacrum). Net average metabolic rate (Brokway1987) was calculated using oxygen and carbon dioxide consumption information obtained from a mobile metabolic system (Oxycon Mobile, CareFusion, San Diego, CA, USA). User preference was taken on a scale from -10 to 10 for both balance and comfort (with 0 being equivalent to walking on a prescribed device, 10 being effortless walking and -10 being very difficult to walk). A detailed calculation method of each balance-related measure is described in (Kim and Collins, 2015b).

Data analysis

Data analysis for each individual: The effect of this controller on each subject was initially evaluated by investigating the single-case experiment results in terms of significance and reliability. This controller was considered to moderately and significantly lower effort if the difference between the Neutral controller condition to the Stabiizing controller condition was higher than 6% and 10%, respectively, based on previous results (Kim and Collins, 2015b). For this comparison, we averaged two Neutral conditions of A-B-A trials. The reliability was first visually examined by investigating the Stabilizing controller condition (B) exhibited lowered efforts compared both the Neutral controller conditions (A). If we observed clear ordering effects, then we further examined reliability by conducting statistical analysis across three days to achieve a semi-random order effect (Edgington, 1967, 1969, 1972). To increase data points (Gottman, 1973), we analyzed responders' (individuals who reduced metabolic rate in the single-case experiment) data by comparing the data from the stabilizing controller to that of the no controller condition. We averaged results of the Stabilizing high controller and Stabilizing low controller on the first two days to represent the stabilizing controller condition. We conducted a three-way ANOVA with significance level of 0.05 to understand the effects of subject, condition, and training. If training is insignificant, we also conducted a repeated measures ANOVA with significance level of 0.05 to understand the effect of subject, and condition.

External validity: The external validity of the controller was conducted by examining group response data when ordering effect exists. We conducted a two-way ANOVA using a linear model with a significance level of 0.05. Once the statistical significance was found, we further conducted a paired t-test by comparing each

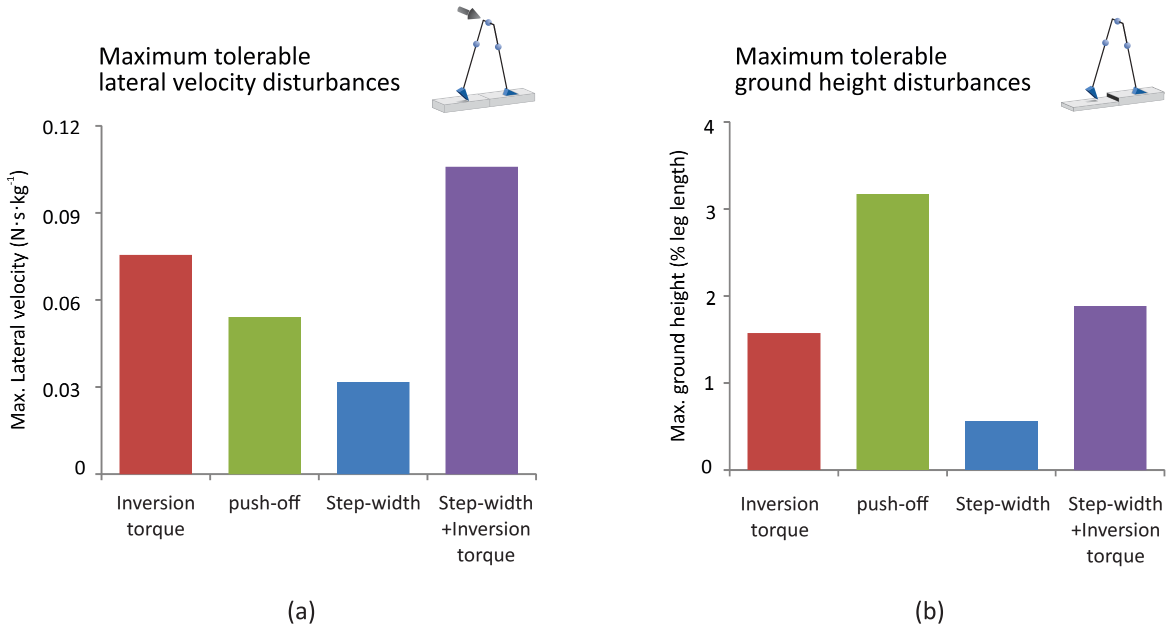

Figure 7.2: (a) Maximum tolerable disturbance in lateral disturbance and (b) maximum tolerable disturbance in ground height. Step-to-step ankle inversion/eversion torque control in combination with step width control increased the ability to restore balance after lateral velocity disturbances. The control method also moderately restored balance from both disturbances. Step-to-step ankle push-off work control was the best method to restore balance after ground height disturbances, and showed a moderate aptitude for restore balance after lateral disturbance.

controller condition to the Destabilizing high gain controller. We also performed a post-hoc power analysis for the condition to estimate the correct sample size.

7.3 Results

7.3.1 Simulation results

Once-per-step ankle inversion/eversion torque control moderately restored balance from disturbances compared to other control methods. The model tolerated 0.075 N·s·kg^{-1} lateral impulse and 1.4% leg length for ground height disturbances (Fig. 7.2). For lateral impulse disturbances, when the inversion/eversion controller was used in conjunction with step-width control, balance restoring performance significantly increased (0.11 N·s·kg^{-1}) compared to step-width control alone (0.032

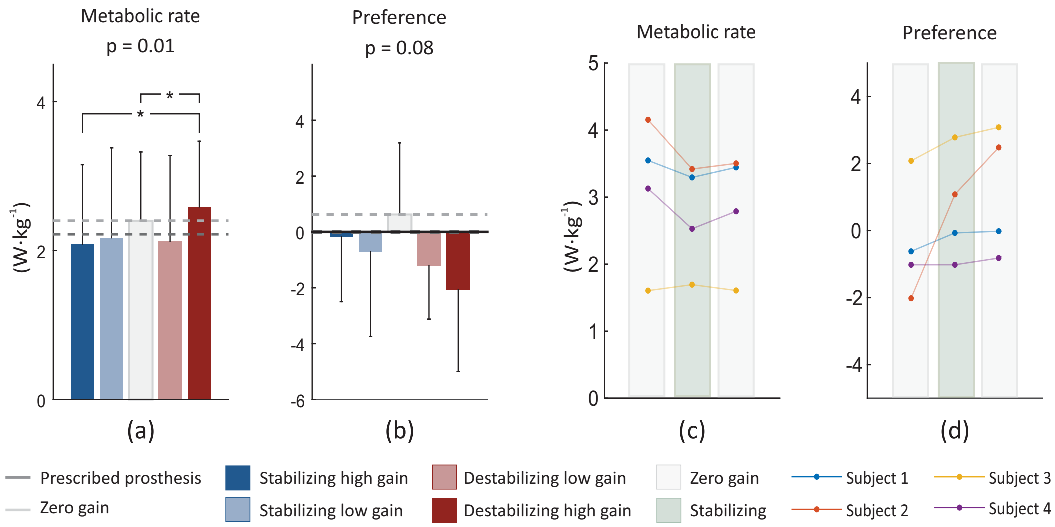

Figure 7.3: Metabolic rate and user preference on day 2 and day 3. On day2, subjects reduced metabolic rate (a) and preferred (b) stabilizing controller condition compared to destabilizing controller condition. On the 3^{rd} day, three subjects reduced metabolic rate (c) compared to the Zero gain controller condition. Clear ordering effect was observed on user preference (d).

0.075 N·s·kg^{-1}, Fig. 7.2 (a)). When ground height disturbances were applied, the inversion/eversion controller was still effective at restoring balance, but the best performance was achieved by the push-off work controller (2.9% leg length, Fig. 7.2 (b)).

7.3.2 Experimental results

Prosthesis mechanics

The controller provided stabilizing/destabilizing inversion/eversion toques while maintaining average torque ($p > 0.5$). The controller tracked the desired inversion/eversion torque within a mean square error of 1 Nm.

Balance-related outcomes

Data analysis for each individual: Three responder subjects (who responded the Stabilizing controller) reduced metabolic rate by 6%, 11%, and 14% for the stabilizing controller condition, respectively. The reduction seemed to consistent across days for each subject when compared to their minimum (p = 0.02, 0.3, and 0.04, respectively). When considering all subject's response across days, their reduction was consistent (p = 0.007). For them, the training effect was not significant (p = 0.8). One subject actually increased his metabolic energy consumption by 5%. For this subject, data from the three days of metabolic rate showed p-value of 0.6 .

Subjects rated the Stabilizing controller and averaged Neutral controller as -0.2 and -0.3, 3.2 and 2.9, 1.1 and -0.3, and -1.0 and -1.0 for balance. Each subject also graded for the Stabilizing controller and averaged neutral controller as -0.1 and -0.3, 2.8 and 2.6, 1.1 and 0.25, and -1 and -0.9 for comfort. However, subjects preferred the second Neutral controller condition compared to the Stabilizing controller condition.

Other measures did not show any trend.

External validity: Metabolic rate was statistically significantly reduced for the Stabilizing controller (ANOVA, p = 0.013) (Fig. 7.3(a)). Metabolic rates were decreased by 19% and 7% for the Stabilizing high controller and the Zero gain controller condition, compared to the Destabilizing high controller condition (p = 0.03 and 0.009, respectively). Achieved powers were 70% and 77%, respectively.

Subjects also showed a trend of preferring Stabilizing controllers in terms of both comfort and balance (ANOVA, p = 0.075 and 0.21, respectively) (Fig. 7.3(b)). The Stabilizing high controller seemed to be favored by 92% and 84% for comfort and balance, compared to the Destabilizing high controller with average scores of -0.4, -2.2, -0.2, and -2.1 for the Stabilizing high comfort, Destabilizing high comfort, Stabilizing high balance, and Stabilizing high comfort, respectively. To avoid type II error with 70% power, 13 subjects and 10 subjects are necessary for the comparison of the Stabilizing high controller and Destabilizing high controller conditions assuming the same distribution for balance and comfort, respectively.

Other measures did not show a trend. The p-values for other outcomes were 0.25, 0.46, and 0.95 for step width variability, average step width, and center of pressure variability in intact limb side, respectively.

7.4 Discussion

In this study, we hypothesized that step-to-step ankle inversion/eversion constant torque modulation would reduce balance-related efforts for some individuals with below knee amputation. We hypothesized that in a simulation study, the controller would show comparable performance to other balance restoring methods. In an subsequent experimental study, we hypothesized some subjects would reduce balance-related effort. We found that using the controller, the model effectively restored balance under disturbances, especially in concert with a foot placement controller. Experimental results also showed that this controller reduced metabolic rate compared to the destabilizing controller. Some subjects even consistently reduced metabolic rate across sessions. This reduction was due to step-to-step changes in inversion/eversion torque, and not average behavior since the ankle inversion/eversion torque was maintained as a desired nominal value,

The observed reductions in metabolic rate, along with the simulation results of disturbance rejection, suggest that this controller can be a possible balance assistance resource. The repeated reduction of the metabolic rate for some subjects further validated this finding. The reductions in metabolic rate also seemed to correlate with user preference, similar to our previous study (Kim and Collins, 2015b,a). In the current study, a stronger trend existed between the reduction in metabolic rate and preference of the stabilizing controller over the destabilizing controller. The positive outcomes in metabolic rate (ANOVA, $p = 0.029$) and tendency toward preference (ANOVA, $p = 0.067$ for balance and 0.052 for comfort) were shown even from the first day(Fig. 7.4). Even though the period was not fully randomized, the consistency of the positive outcomes on the stabilizing controllers over destabilizing controllers suggests that this controller is a possible balance restoring resource.

The correlation of metabolic rate and user preference in this study seems to be a result of transparent balance assistance behavior. Similar outcomes were observed in our previous push-off work study, but the statistical significance was negligible (Kim and Collins, 2015b). This might be due to complicated device behavior - lateral motion was stabilized by affecting sagittal motion (Kim and Collins, 2014). The intricate push-off work controller seemed to be understood through forced exploration period, shown by an increase in statistical significance of the user preference. In contrast, another previous study showed that subjects seemed to directly perceive benefit when straightforward controllers were presented (Kim et al., 2015). The spring like controller in ankle inversion/eversion was clearly linked to a strong preference. The action of the step-to-step ankle inversion controller also seemed to be fairly easy to understand from the beginning since it applied *lateral* force based on *lateral* acceleration of the subject. This could explain why the controller significantly impacted user preference and reduced metabolic energy consumption without forced exploration period.

Perhaps, this transparent device behavior enabled the subjects to use other balance restoring resources. Unlike previous studies, we did not see a trend in foot placement (Kim and Collins, 2015b) or intact limb control effort (Kim and Collins, 2015a) during the group experiment period. Instead, the trend existed during the acclimation day. During the acclimation period, subjects showed the following trends: reduced step width variability, reduced step width average, reduced center of pressure variability, and reduced metabolic rate ($p = 0.20$, 0.38, 0.25, and 0.03, respectively) (Fig. 7.4). Perhaps, after acclimation, subjects found and used other balance restoring methods, such as upper body motion (Beurskens et al., 2014; Curtze et al., 2011). By measuring other possible balance resources, we may be able to explain the cause of reduction in metabolic rate.

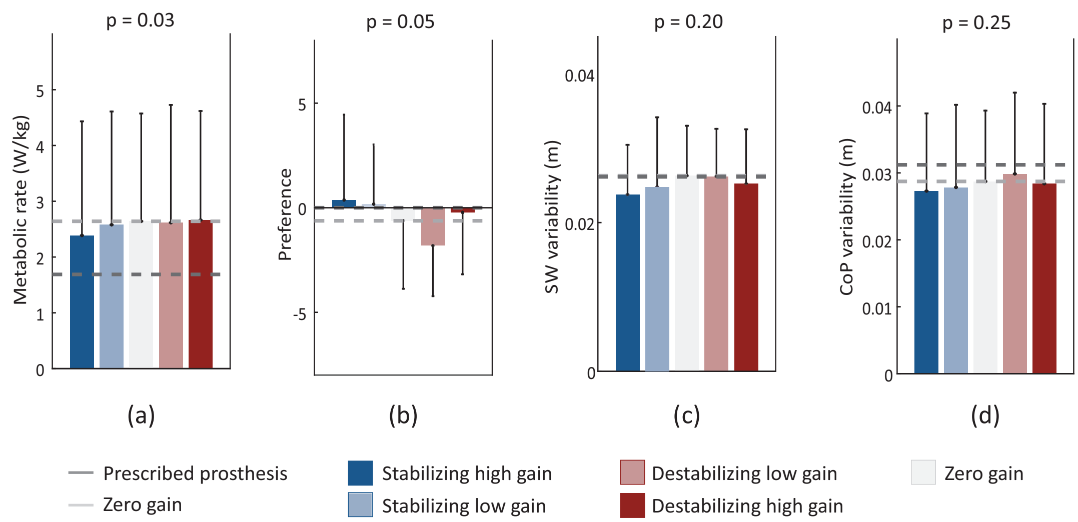

Figure 7.4: Acclimation day results: subjects reduced metabolic rate for the stabilizing controller condition (a) and preferred the condition (b). They showed a trend of reducing step width variability (c) and intact limb control effort (e).

The interpretation of this study, though, is limited due to the small number of subjects. Amputees have more inter-subject variability than their able-bodied counterparts (Fey et al., 2010; Wentink et al., 2013). We also experienced additional variability due to age (Nagano et al., 2013; Kadono and Pavol, 2013), compared to our previous study (Kim and Collins, 2015b). This could possibly be compensated by comparing the Stabilizing high controller to the Destabilizing high controller, such that the difference between the conditions was largest. However, when the effect size became small, larger number of subject was necessary. By conducting experiments with more subjects, the effect of this controller on balance-related effort might be more clearly understood.

In this study, subjects did not seem to benefit from the forced exploration period. The observed metabolic reduction was not statistically significantly different and subjects did not appear to reduce foot placement or intact limb control effort. They still preferred the neutral controller after forced exploration. Intervention can be differently learned by experimental design (Selinger et al., 2015). By carefully improving our current experimental design, subjects may be able to discover a way to use the proposed controller more effectively.

Our simulation study suggests that ankle inversion/eversion torque control was better able to handle specific types of disturbances, compared to the push-off work controller. Both controllers showed a potential to reduce balance-related effort (Kim and Collins, 2015a,b). Perhaps an appropriate combination of these controllers might further enhance balance to withstand several disturbances, including a combination of lateral velocity and ground height disturbances.

This study showed that ankle inversion/eversion control has the potential to reduce balance-related effort, as shown by comparable disturbance tolerance in our simulation study and reduction in metabolic rate in our experimental study. The benefit of this controller might be further uncovered by conducting experiments with more subjects.

Chapter 8

Conclusions

Individuals with below knee amputation exert more balance-related effort (Paysant et al., 2006) and would prefer to have a balance assisting device (Legro et al., 1999). Despite great progress in ankle-foot prostheses (Herr and Grabowski, 2012; Zelik et al., 2011), the balance assisting capability of such devices has been inconclusive (Gates et al., 2013b). Motivated by this unclear assistance in balance, I developed controllers to improve balance for individuals with below knee amputation using a dynamic walking model.

To develop balance assistive controllers, I first investigated the effectiveness of ankle actuation on stability by comparing the control technique to active foot placement control. The comparison was done using a three dimensional walking model of an amputees gait. This type of model can capture basic dynamics (Kuo, 1999; Donelan et al., 2001) and can help motivate the design of potential controller (Hobbelen et al., 2008; Bhounsule et al., 2014). In addition, a simulation study can act as a rigorous platform on which to compare control techniques. Using this approach, I revealed that ankle actuation, especially ankle push-off work modulation, is as important as foot placement control at restoring balance, regardless of the type of disturbance and speed of model. Even though the importance of ankle actuation in balance was previously considered in two dimensional walking (Hobbelen et al., 2008; Bhounsule et al., 2014), simulating with a three dimensional model of limit cycle walking further confirmed the potential benefit of using active step-to-step modulation in ankle actuation as a balance assisting resource for individuals with below knee amputation.

Motivated by the simulation result of ankle push-off work modulation on balance, I explored the benefit of the controller on balance during locomotion by conducting a human subject experiment on individuals with simulated amputation. Modulation of ankle push-off work once per step resulted in a reduction in metabolic rate. This reduction might have been achieved by reducing foot placement control effort (O'Connor et al., 2012). This finding demonstrates that it is possible to reduce walking effort by providing balance assistance with an active device. In addition, because average push-off work was held approximately constant across trials, it may be possible for this controller to be realized without additional high power actuation. This promising outcome further inspired the development of an emulator with an additional degree of freedom and a balance perturbing device.

We more fully confirmed the efficacy of this controller by conducting an experiment with the target population - individuals with below knee amputation. We designed the experiment (Dermer and Hoch, 2012) to account for high inter-subject variability (Fey et al., 2010; Wentink et al., 2013). Additionally, we incorporated a forced exploration period (Selinger et al., 2015) to help participants better understand the behavior of the device. After forced exploration, participants reduced their intact limb control effort during single support. Some subjects exhibited reduction in their metabolic rate and they changed their opinion on the stabilizing controller in a favorable way. This result shows that the target population can also reduce balance-related effort using a balance assisting device. In addition, the importance of training also emerged (Selinger et al., 2015). In this particular case, the training impacted not only energy consumption but also preference.

The simulation study and experimental study also inspired the exploration of another potential balance assistance technique - ankle inversion/eversion torque control. In order to be able to test ideas related to ankle inversion/eversion torque control, we developed an ankle-foot prosthesis with two independently-actuated toes. With this configuration this device is capable of providing ankle inversion/ eversion and ankle plantarflexion. In addition to the added degree of freedom, this emulator outperformed the previous device, partially due to improved sensing. As a result, this device allowed for the development of control ideas with fewer constraints and also influenced the design of the sensory systems used in our lab (Witte et al., 2015).

Using this two Degrees of Freedom emulator, we examined the effect of ankle inversion/eversion stiffness on balance by conducting an experiment on individuals with below knee amputation. The ankle-foot prosthesis applied a restoring inversion torque at each step by multiplying a stiffness by the ankle inversion angle deviation from a nominal value during stance while maintaining zero average torque. Subjects walked under different conditions in which the device provided either a restoring torque, a destabilizing torque, or neither. Subjects reduced foot placement and intact limb control effort for the restoring torque condition, compared to destabilizing torque condition. This result suggests that ankle inversion/eversion stiffness can affect balance. However, they did not reduce their metabolic rate, possibly due to lack of sufficient training (Selinger et al., 2015).

Based on the prior result, we designed a controller that delivered constant ankle inversion/eversion torque at each step, proportional to the deviation of the lateral acceleration from the nominal acceleration. Then, we performed a human subject experiment on individuals with below knee amputation. We also included an exploration period to give additional training. We found that subjects reduced their metabolic rate compared to the destabilizing controller starting on the first day. We thought this reduction may be due to the transparency of the controllers behavior, i.e. providing lateral torque based on a deviation in lateral acceleration. Energy reduction seemed to be a result of a decrease in balance-related effort that was not related to foot placement (O'Connor et al., 2012) or intact limb center-of-pressure variability (Hof et al., 2007). Experimenting with more subjects while collecting other possible balance-related measures, such as arm sway, may more clearly reveal the effect of this inversion/eversion controller.

Overall, I demonstrated that controlling ankle actuation, especially on a step-to-step basis, has the potential to reduce individuals balance-related effort. While on average people reduced balance-related effort when presented with stabilizing controllers, we observed variation in the response of each individual. This variability suggests that customizing this type of controller to each person may result in a greater reduction in balance-related effort. One specific path for future research could be individualizing the controller using online optimization (Koller et al., 2015; Felt et al., 2014; Caputo et al., 2015b).

Instead of individualizing each controller, we might also consider helping participants adapt to the controller (Selinger et al., 2015). In other words, it may be useful to consider the complex human-device interaction that exists when designing a training method or device optimization technique. In our experimental study, we observed that some controllers were effective only after subjects completed a forced exploration session. This result suggest that suitable instruction may help subjects better learn and understand the controller, and may improve the controller's efficacy. It may be important to allow for the human to first adapt the behavior of the device before the device is optimized.

Through the conduction of the above-stated research, another possible balance-assistance technique became apparent: controlling fore-aft motion by using an active heel. By employing such actuation, we may be able to provide soft heel contact and control the location of the center of pressure. Intact limb center of pressure control seemed to be used especially among individuals with below knee amputation (Hof et al., 2007). Future work could control such features and thus further contribute to decrease their balance-related effort.

This thesis exhibited the importance of active ankle control for reducing balance-related effort in individuals with below knee amputation, initiated by results of a simple dynamic walking model. While this study is limited by a small number of participants, future work could generalize this finding. In addition, the controllers presented in this dissertation executed step-to-step modification without changing average behavior. Therefore, implementing this type controller in an active commercial device might not need additional actuation. By incorporating this controller in commercially-available prostheses, individuals with below knee amputation may be able to improve their mobility and quality of life.

Bibliography

Abul-Haj, C. and Hogan, N. (1987). An emulator system for developing improved elbow-prosthesis designs. *IEEE Transactions on Biomedical Engineering*, 9:724–737.

Adamczyk, P. G., Roland, M., and Hahn, M. E. (2013). Novel method to evaluate angular stiffness of prosthetic feet from linear compression tests. *ASME Journal of Biomechanical Engineering*, 135:104502.

Adams, P. F., Hendershot, G. E., and Marano, M. A. (1999). Current estimates from the national health interview survey, 1996. *Vital and health statistics. Series 10, Data from the National Health Survey*, (200):1–203.

Agarwal, G. C. and Gottlieb, G. L. (1977). Oscillation of the human ankle joint in response to applied sinusoidal torque on the foot. *Journal of Physiology*, 268:151–176.

Alcaide-Aguirre, R. E., Morgenroth, D. C., and Ferris, D. P. (2013). Motor control and learning with lower-limb myoelectric control in amputees. *Journal of Rehabilitation Research & Development*, 50(5):687–698.

Aldridge, J. M., Sturdy, J. T., and Wilken, J. M. (2012). Stair ascent kinematics and kinetics with a powered lower leg system following transtibial amputation. *Gait & Posture*, 36(2):291–5.

Andersen, J. B. and Sinkjær, T. (1995). An actuator system for investigating electrophysiological and biomechanical features around the human ankle joint during gait. *IEEE Transactions on Rehabilitation Engineering*, 3(4):299–306.

Au, S. K. and Herr, H. (2009). Powered ankle-foot prosthesis improves walking metabolic economy. *IEEE Transactions on Robotics*, 25:51–66.

Au, S. K., Weber, J., and Herr, H. (2007). Biomechanical design of a powered ankle-foot prosthesis. In *Proceedings of the IEEE International Conference on Rehabilitation Robotics (ICORR)*, pages 298–303.

Bauby, C. E. and Kuo, A. D. (2000). Active control of lateral balance in human walking. *Journal of Biomechanics*, 33:1433–1440.

Bawa, P. and Stein, R. B. (1976). Frequency response of human soleus muscle. *Journal of Neurophysiology*, 39(4):788–793.

Bellman, R. D., Holgate, M. A., and Sugar, T. G. (2008). SPARKy 3: Design of an active robotic ankle prosthesis with two actuated degrees of freedom using regenerative kinetics. In *Proceedings of the IEEE International Conference on Biomedical Robotics and Biomechatronics (BioRob)*, pages 511–516.

Beltran, E. J., Dingwell, J. B., and Wilken, J. M. (2014). Margins of stability in young adults with traumatic transtibial amputation walking in destabilizing environments. *JBiomech*, 47(5):1138–1143.

Benallegue, M., Laumond, J.-P., and Berthoz, A. (2013). Contribution of actuated head and trunk to passive walkers stabilization. In *Proceedings of the IEEE International Conference on Robotics and Automation (ICRA)*, pages 5638–5643.

Beurskens, R., Wilken, J. M., and Dingwell, J. B. (2014). Dynamic stability of individuals with transtibial amputation walking in destabilizing environments. *Journal of biomechanics*, 47(7):1675–1681.

Bhounsule, P., Cortell, J., and Ruina, A. (2012). Cornell ranger: Implementing energy-optimal trajectory control using low information, reflex-based control. In *Proceedings of Dynamic Walking*.

Bhounsule, P. A., Cortell, J., Grewal, A., Hendriksen, B., Karssen, J. D., Paul, C., and Ruina, A. (2014). Low-bandwidth reflex-based control for lower power walking: 65 km on a single battery charge. *International Journal of Robotics Research*, 33(10):1305–1321.

Bisi, M. C., Stagni, R., Houdijk, H., and Gnudi, G. (2011). An EMG-driven model applied for predicting metabolic energy consumption during movement. *Journal of Electromyography and Kinesiology*, 21:1074–1080.

Blake, O. M. and Wakeling, J. M. (2013). Estimating changes in metabolic power from EMG. *SpringerPlus*, 2:229.

Bouwsema, H., van der Sluis, C. K., and Bongers, R. M. (2010). Learning to control opening and closing a myoelectric hand. *Archives of physical medicine and rehabilitation*, 91(9):1442–6.

Box, G. (2012). *Box and Jenkins: time series analysis, forecasting and control*. A Very British Affair: Six Britons and the Development of Time Series Analysis During the 20th Century.

Brockway, J. M. (1987). Derivation of formulae used to calculate energy expenditure in man. *Human Nutrition: Clinical Nutrition*, 41C:463–471.

Bruijn, S., Meijer, O., Beek, P., and Van Dieën, J. (2013). Assessing the stability of human locomotion: a review of current measures. *JRSI*, 10(83):20120999.

Bruijn, S. M., Meijer, O. G., Beek, P. J., and van Dieën, J. H. (2010). The effects of arm swing on human gait stability. *Journal of Experimental Biology*, 213:3945–3952.

Caputo, J. M. (2015). *Informing ankle-foot prosthesis design and prescription through systematic experimentation with a tethered robotic prosthesis*. PhD dissertation, Carnegie Mellon University, Mechanical Engineering. Advisor: Steven H. Collins.

Caputo, J. M., Adamczyk, P. G., and Collins, S. H. (2015a). Informing ankle-foot prosthesis prescription through haptic emulation of candidate devices. In *Proceedings of the IEEE International Conference on Robotics and Automation (ICRA)*, pages 6445 – 6450.

Caputo, J. M., Adamczyk, P. G., and Collins, S. H. (2015b). Optimizing prosthesis design to maximize user satisfaction using a tethered robotic ankle-foot prosthesis. In *Proceedings of Dynamic Walking*.

Caputo, J. M. and Collins, S. H. (2013). An experimental robotic testbed for accelerated development of ankle prostheses. In *Proceedings of the IEEE International Conference on Robotics and Automation (ICRA)*, pages 2630–2635.

Caputo, J. M. and Collins, S. H. (2014a). Prosthetic ankle push-off work reduces metabolic rate but not collision work in non-amputee walking. *Nature Scientific Reports*, 4:7213.

Caputo, J. M. and Collins, S. H. C. (2014b). A universal ankle-foot prosthesis emulator for experiments during human locomotion. *ASME Journal of Biomechanical Engineering*, 136:035002.

Chandler, R., Clauser, C., McConville, J., Reynolds, H., and Young, J. (1975). Investigation of inertial properties of the human body. Technical report, Air Force Aerospace Medical Research Lab Wright-Patterson AFB OH. No. AMRL-TR-74-137.

Cherelle, P., Grosu, V., Matthys, A., Vanderborght, B., and Lefeber, D. (2014a). Design and validation of the ankle mimicking prosthetic (AMP-) foot 2.0. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 22(1):138–148.

Cherelle, P., Mathijssen, G., Wang, Q., Vanderborght, B., and Lefeber, D. (2014b). Advances in propulsive bionic feet and their actuation principles. *Adv. Mech. Eng.*, 6:984046.

Childers, W. L., Funderburk, R. D., Smith, A. T., and Davidson, C. J. (2015). The effect of muli-axial prosthetic stiffness on angular momentum in people with transtibial amputation walking over uneven terrain. In *Proceedings of the American Society of Biomechanics Annual Meeting*, page 185, Columbus, Ohio.

College Park. College Park - Trustep. September, 2014. URL: http://www.college-park.com/prosthetics/trustep.

Collins, S. H. (2013). What do walking humans want from mechatronics? In *In Proceedings of the IEEE International Conference on Mechatronics (ICM)*, pages 24–27.

Collins, S. H., Adamczyk, P. G., and Kuo, A. D. (2009). Dynamic arm swinging in human walking. *Proceedings of the Royal Society of London B: Biological Sciences*, 276:3679–3688.

Collins, S. H., Caputo, J. M., and Adamczyk, P. G. (2014). Emulating prosthetic feet during the prescription process to improve outcomes and justifications. In *Proceedings of the IEEE Workshop on Advanced Robotics and its Social Impacts (ARSO)*, pages 1–2.

Collins, S. H. and Jackson, R. W. (2013). Inducing self-selected human engagement in robotic locomotion training. In *Proceedings of the IEEE International Conference on Rehabilitation Robotics (ICORR)*, pages 1–6.

Collins, S. H., Kim, M., Chen, T., and Chen, T. (2015a). An ankle-foot prosthesis emulator with control of plantarflexion and inversion-eversion torque. In *Proceedings of the IEEE International Conference on Robotics and Automation (ICRA)*, pages 1210 – 1216.

Collins, S. H. and Kuo, A. D. (2013). Two independent contributions to step variability during over-ground human walking. *Public Library of Science: ONE*, 8(8):e73597.

Collins, S. H., Wiggin, M. B., and Sawicki, G. S. (2015b). An exoskeleton that uses no energy yet reduces the metabolic cost of human walking. *Nature*, 522:212–215.

Crenshaw, J. R., Kaufman, K. R., and Grabiner, M. D. (2013a). Compensatory-step training of healthy, mobile people with unilateral, transfemoral or knee disarticulation amputations: A potential intervention for trip-related falls. *Gait & Posture*, 38(3):500–6.

Crenshaw, J. R., Kaufman, K. R., and Grabiner, M. D. (2013b). Trip recoveries of people with unilateral, transfemoral or knee disarticulation amputations: Initial findings. *Gait & Posture*, 38:534–536.

Curtze, C., Hof, A. L., Postema, K., and Otten, B. (2011). Over rough and smooth: amputee gait on an irregular surface. *Gait & posture*, 33(2):292–6.

Dean, J. C., Alexander, N. B., and Kuo, A. D. (2007). The effect of lateral stabilization on walking in young and old adults. *IEEE Transactions on Biomedical Engineering*, 54(11):1919–1926.

Dermer, M. L. and Hoch, T. A. (2012). Improving descriptions of single-subject experiments in research texts written for undergraduates. *The Psychological Record*, 49(1):49–66.

Dingwell, J. B., Cusumano, J. P., Cavanagh, P. R., and Sternad, D. (2001). Local dynamic stability versus kinematic variability of continuous overground and treadmill walking. *ASME Journal of Biomechanical Engineering*, 123(1):27–32.

Dollar, A. M. and Herr, H. (2008). Lower extremity exoskeletons and active orthoses: challenges and state-of-the-art. *IEEE Transactions on Robotics*, 24(1):144–158.

Donelan, J. M., Kram, R., and Kuo, A. D. (2001). Mechanical and metabolic determinants of the preferred step width in human walking. *Proceedings of the Royal Society of London B: Biological Sciences*, 268:1985–1992.

Donelan, J. M., Shipman, D. W., Kram, R., and Kuo, A. D. (2004). Mechanical and metabolic requirements for active lateral stabilization in human walking. *Journal of Biomechanics*, 37(6):827–35.

Edgington, E. S. (1967). Statistical inference from n = 1 experiments. *Journal of Psychology*, 65(2):195–199.

Edgington, E. S. (1969). *Statistical inference: The distribution-free approach*. McGraw-Hill, New York.

Edgington, E. S. (1972). The design of one-subject experiments for testing hypotheses. *Western Psychologist*, 3:33–38.

Eng, J. J. and Winter, D. A. (1995). Kinetic analysis of the lower limbs during walking: what information can be gained from a three-dimensional model? *Journal of Biomechanics*, 28(6):753–758.

Espy, D. D., Yang, F., Bhatt, T., and Pai, Y.-C. (2010). Independent influence of gait speed and step length on stability and fall risk. *Gait & posture*, 32(3):378–82.

Esquenazi, A. and DiGiacomo, R. (2001). Rehabilitation after amputation. *Journal of the American Podiatric Medical Association*, 91(1):13–22.

Fabre, J. M., Ellis, R., Kosma, M., and Wood, R. H. (2010). Falls risk factors and a compendium of falls risk screening instruments. *Journal of Geriatric Physical Therapy*.

Felt, W., Selinger, J., Donelan, J. M., and Remy, C. D. (2014). Body-in-the-loop — optimizing actual human walking. In *Proceedings of Dynamic Walking*, pages 1–2.

Ferris, D. P., Gordon, K. E., and Sawicki, G. S. (2006). An improved powered ankle-foot orthosis using proportional myoelectric control. *Gait & Posture*, 23:425–428.

Fey, N., Silverman, A., and Neptune, R. (2010). The influence of increasing steady-state walking speed on muscle activity in below-knee amputees. *Journal of Electromyography and Kinesiology*, 20(1):155–161.

Ficanha, E. M., Rastgaar, M., Moridian, B., and Mahmoudian, N. (2013). Ankle angles during step turn and straight walk: Implications for the design of a steerable ankle-foot prosthetic robot. In *Proceedings of the ASME Dynamic Systems and Control Conference (DSCC)*, pages 1–5.

Fu, Q. and Santello, M. (2015). Retention and interference of learned dexterous manipulation: interaction between multiple sensorimotor processes. *Journal of Neurophysiology*, 113(1):144–155.

Gates, D. H., Aldridge, J. M., and Wilken, J. M. (2013a). Kinematic comparison of walking on uneven ground using powered and unpowered prostheses. *ClinBiom*, 28(4):467–472.

Gates, D. H., Scott, S. J., Wilken, J. M., and Dingwell, J. B. (2013b). Frontal plane dynamic margins of stability in individuals with and without transtibial amputation walking on a loose rock surface. 38(4):570–575.

Geertzen, J. H. B., Bosmans, J. C., van der Schans, C. P., and Dijkstra, P. U. (2005). Claimed walking distance of lower limb amputees. *Disability and rehabilitation*, 27(3):101–104.

Gentile, J. Ronald, A. H. R. and Klein, R. D. (1972). An analysis-of-variance model for the intra subject replication design. *Journal of Applied Behavior Analysis*, 5(2):193.

Goldfarb, M., Lawson, B. E., and Schultz, A. H. (2013). Realizing the promise of robotic leg prostheses. *Science Translational Medicine*, 5(210):1–6.

Golenia, L., Schoemaker, M. M., Mouton, L. J., and Bongers, R. M. (2014). Individual Differences in Learning a Novel Discrete Motor Task. *PLoS ONE*, 9(11):e112806.

Gonzalez, E. and Mathews, M. (1980). Femoral fractures in patients with lower extremity amputations. *PMR*, 61(6):276–280.

Gooday, H. M. and Hunter, J. (2004). Preventing falls and stump injuries in lower limb amputees during inpatient rehabilitation: completion of the audit cycle. *CR*, 18(4):379–390.

Gordon, K. E., Sawicki, G. S., and Ferris, D. P. (2006). Mechanical performance of artificial pneumatic muscles to power an ankle-foot orthosis. *Journal of Biomechanics*, 39:1832–1841.

Gottman, J. M. (1973). N-of-one and n-of-two research in psychotherapy. *Psychological Bulletin*, 80(2):93.

Griffiths, P. G., Gillespie, R. B., and Freudenberg, J. S. (2011). A fundamental linear systems conflict between performance and passivity in haptic rendering. *IEEE Transactions on Robotics*, 27(1):75–88.

Hafner, B. J., Sanders, J. E., Czerniecki, J. M., and Fergason, J. (2002). Energy storage and return prostheses: does patient perception correlate with biomechanical analysis? *Clinical Biomechanics*, 17(5):325–44.

Hagberg, K. and Brånemark, R. (2001). Consequences of non-vascular trans-femoral amputation: a survey of quality of life, prosthesis use and problems. *Prosthetics and Orthotics International*, 25:186–194.

Hannaford, B. and Okamura, A. (2008). Haptics. In *Springer Handbook of Robotics*, pages 719–739, Heidelberg, Germany. Springer Berlin.

Hansen, A. H., Childress, D. S., and Knox, E. H. (2004). Roll-over shapes of human locomotor systems: effects of walking speed. *Current Biology*, 19:407–414.

Hansen, N. (2006). The cma evolution strategy: A comparing review. In Lozano, J. A., editor, *Towards a New Evolutionary Computation*, pages 75–102. Springer, Berlin.

Hansen, N. and Kern, S. (2004). Evaluating the CMA evolution strategy on multimodal test functions. In Yao, X. et al., editors, *Parallel Problem Solving from Nature PPSN VIII*, volume 3242 of *LNCS*, pages 282–291. Springer.

Hartmann, D. P. (1974). Forcing square pegs into round holes: some comments on "an analysis of variance model for the intra subject replication design".

Hausdorff, J. M., Rios, D. A., and Edelberg, H. K. (2001). Gait variability and fall risk in community-living older adults: a 1-year prospective study. *Archives of Physical Medicine and Rehabilitation*, 82(8):1050–1056.

Hawes, M. R. and Sovak, D. (1994). Quantitative morphology of the human foot in a North American population. *Ergonomics*, 37(7):1213–1226.

Heidi Nadollek, S. B. and Isles, R. (2002). Outcomes after trans-tibial amputation: the relationship between quiet stance ability, strength of hip abductor muscles and gait. 7(4):203–214.

Herr, H. M. and Grabowski, A. M. (2012). Bionic ankle-foot prosthesis normalizes walking gait for persons with leg amputation. *Proceedings of the Royal Society of London B: Biological Sciences*, 279:457–464.

Highsmith, M. J., Kahle, J. T., Bongiorni, D. R., Sutton, B. S., Groer, S., and Kaufman, K. R. (2010). Safety, energy efficiency, and cost efficacy of the C-Leg for transfemoral amputees: A review of the literature. *Prosthetics and Orthotics International*, 34(4):362–377.

Hitt, J., Sugar, T., Holgate, M., Bellman, R., and Hollander, K. (2009). Robotic transtibital prosthesis with biomechanical energy regeneration. *Industrial Robot: An International Journal*, 36(5):441–447.

Hitt, J. K., Sugar, T. G., Holgate, M., and Bellman, R. (2010). An active foot-ankle prosthesis with biomechanical energy regeneration. *Journal of Medical Devices*, 4.

Hobbelen, D., de Boer, T., and Wisse, M. (2008). System overview of bipedal robots flame and tulip: Tailor-made for limit cycle walking. In *Proceedings of the IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*, pages 2486–2491.

Hobbelen, D. G. E. and Wisse, M. (2007). A disturbance rejection measure for limit cycle walkers: The gait sensitivity norm. *IEEE Transactions on Robotics*, 23:1213–1224.

Hobbelen, D. G. E. and Wisse, M. (2008). Ankle actuation for limit cycle walkers. *International Journal of Robotics Research*, 27(6):709–735.

Hof, A. L., Gazendam, M. G. J., and Sinke, W. E. (2005). The condition for dynamic stability. *Journal of Biomechanics*, 38(1):1–8.

Hof, A. L., van Bockel, R. M., Schoppen, T., and Postema, K. (2007). Control of lateral balance in walking: Experimental findings in normal subjects and above-knee amputees. *Gait & Posture*, 25:250–258.

Hof, A. L., Vermerris, S. M., and Gjaltema, W. A. (2010). Balance responses to lateral perturbations in human treadmill walking. *Journal of Experimental Biology*, 213(15):2655–2664.

Hogan, N. (1984). Impedance control: An approach to manipulation. In *American Controls Conference*, pages 304–313.

Hollman, J. H., Kovash, F. M., Kubik, J. J., and Linbo, R. A. (2007). Age-related differences in spatiotemporal markers of gait stability during dual task walking. *Gait & Posture*, 26(1):113–119.

Horak, F. B., Shupert, C. L., and Mirka, A. (1989). Components of postural dyscontrol in the elderly: a review. *Neurobiology of Aging*, 10(6):727–738.

Hsu, M. J., Nielsen, D. H., Lin-Chan, S. J., and Shurr, D. (2006). The effects of prosthetic foot design on physiologic measurements, self-selected walking velocity, and physical activity in people with transtibial amputation. *Archives of Physical Medicine and Rehabilitation*, 87:123–129.

Huang, S. and Ferris, D. P. (2012). Muscle activation patterns during walking from transtibial amputees recorded within the residual limb-prosthetic interface. *Journal of neuroengineering and rehabilitation*, 9(1):55.

Huang, S., Wensman, J. P., and Ferris, D. P. (2014). An experimental powered lower limb prosthesis using proportional myoelectric control. *Journal of Medical Devices*, 8(2):024501.

Hunt, A. E., Smith, R. M., and Torode, M. (2001). Extrinsic muscle activity, foot motion and ankle joint moments during the stance phase of walking. *Foot & Ankle*, 22(1):31–41.

Hurmuzlu, Y. and Moskowitz, G. D. (1986). Role of impact in the stability of bipedal locomotion. *Dynamics and Stability of Systems*, 1(3):217–234.

Hurt, C. P., Rosenblatt, N., Crenshaw, J. R., and Grabiner, M. D. (2010). Variation in trunk kinematics influences variation in step width during treadmill walking by older and younger adults. *Gait & Posture*, 31(4):461–464.

IJmker, T., Houdijk, H., Lamoth, C. J., Jarbandhan, A. V., Rijntjes, D., Beek, P. J., and van der Woude, L. H. (2013). Effect of balance support on the energy cost of walking after stroke. *Archives of Physical Medicine and Rehabilitation*, 94(11):2255–61.

Ijmker, T. and Lamoth, C. J. C. (2012). Gait and cognition: the relationship between gait stability and variability with executive function in persons with and without dementia. *Gait & Posture*, 35(1):126–30.

IJmker, T., Noten, S., Lamoth, C. J., Beek, P. J., van der Woude, L. H. V., and Houdijk, H. (2014). Can external lateral stabilization reduce the energy cost of walking in persons with a lower limb amputation? 40(4):616–621.

Jackson, R. W. and Collins, S. H. (2015). An experimental comparison of the relative benefits of work and torque assistance in ankle exoskeletons. *Journal of Applied Physiology*, 119(5):541–557.

Jones, Richard R., R. S. V. and Weinrott, M. (1977). Time series analysis in operant research 1. *Journal of Applied Behavior Analysis*, 10(1):151–166.

Kadono, N. and Pavol, M. J. (2013). Effects of aging-related losses in strength on the ability to recover from a backward balance loss. *Journal of biomechanics*, 46(1):13–18.

Kajita, S., Kanehiro, F., Kaneko, K., Fujiwara, K., Harada, K., Yokoi, K., and Hirukawa, H. (2003). Biped walking pattern generation by using preview control of zero-moment point. In *Proceedings of the IEEE International Conference on Robotics and Automation (ICRA)*, pages 1620–1626.

Kajita, S. and Tan, K. (1991). Study of dynamic biped locomotion on rugged terrain - derivation and application of the linear inverted pendulum mode. In *Proceedings of the IEEE International Conference on Robotics and Automation (ICRA)*, pages 1405–1411.

Kaufman, K. R., Levine, J. a., Brey, R. H., Iverson, B. K., McCrady, S. K., Padgett, D. J., and Joyner, M. J. (2007). Gait and balance of transfemoral amputees using passive mechanical and microprocessor-controlled prosthetic knees. *Gait & Posture*, 26(4):489–493.

Kaufman, K. R., Wyatt, M. P., Sessoms, P. H., and Grabiner, M. D. (2014). Task-specific fall prevention training is effective for warfighters with transtibial amputations. *Clinical Orthopaedics and Related Research*, 472:3076–3084.

Kazdin, A. E. (1975). Characteristics and trends in applied behavior analysis 1. *Journal of Applied Behavior Analysis*, 8(3):332.

Kazdin, A. E. (2012). *Behavior modification in applied settings.* Waveland Press.

Kim, J.-Y., Park, I.-W., and Oh, J.-H. (2007). Walking control algorithm of biped humanoid robot on uneven and inclined floor. *Journal of Intelligent and Robotic Systems*, 48(4):457–484.

Kim, M., Chen, T., Chen, T., and Collins, S. H. CAD models and bill of materials. August, 2015. URL: biomechatronics.cit.cmu.edu/publications/Kim_2015_TRO—CAD_Files.zip.

Kim, M. and Collins, S. H. (2013). Stabilization of a three-dimensional limit cycle walking model through step-to-step ankle control. In *Proceedings of the IEEE International Conference on Rehabilitation Robotics (ICORR)*, pages 1–6.

Kim, M. and Collins, S. H. (2014). Once-per-step control of ankle push-off work improves balance in a three-dimensional simulation of bipedal walking. in review.

Kim, M. and Collins, S. H. (2015a). The effect of ankle push-off work modulation in an ankle-foot prostheis on balance-related effort. in preparation.

Kim, M. and Collins, S. H. (2015b). Once-per-step control of ankle-foot prosthesis push-off work reduces effort associated with balance during walking. *Journal of NeuroEngineering and Rehabilitation*, 12(1):43.

Kim, M., Lyness, H., Chen, T., , and Collins, S. H. (2015). The influence of inversion-eversion stiffness of an ankle-foot prostheses on balance-related effort during walking. in preparation.

Koller, J. R., Jacobs, D. A., Ferris, D. P., and Remy, C. D. (2015). Learning to walk with an adaptive gain proportional myoelectric controller for a robotic ankle exoskeleton. *Journal of NeuroEngineering and Rehabilitation*, 12(1):97.

Krasovsky, T., Lamontagne, A., Feldman, A. G., and Levin, M. F. (2014). Effects of walking speed on gait stability and interlimb coordination in younger and older adults. *Gait & posture*, 39(1):378–385.

Kulkarni, J., Wright, S., Toole, C., Morris, J., and Hirons, R. (1996). Falls in patients with lower limb amputations: prevalence and contributing factors. *PT*, 82(2):130–136.

Kuo, A. D. (1999). Stabilization of lateral motion in passive dynamic walking. *International Journal of Robotics Research*, 18(9):917–930.

Kuo, A. D. (2001). A simple model predicts the step length-speed relationship in human walking. *ASME Journal of Biomechanical Engineering*, 123:264–269.

Kuo, A. D. (2012). The dynamics workbench. http://www-personal.umich.edu/~artkuo/DynamicsWorkbench/.

Kuo, A. D., Donelan, J. M., and Ruina, A. (2005). Energetic consequences of walking like an inverted pendulum: step-to-step transitions. *Exercise and Sport Sciences Reviews*, 33:88–97.

Lamoth, C. J. C., Ainsworth, E., Polomski, W., and Houdijk, H. (2010). Variability and stability analysis of walking of transfemoral amputees. *Medical Engineering and Physics*, 32:1009–1014.

Lawson, B. E., Varol, H. A., Huff, A., Erdemir, E., and Goldfarb, M. (2013). Control of stair ascent and descent with a powered transfemoral prosthesis. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 21:466–473.

Legro, M. W., Reiber, G., del Aguila, M., Ajax, M. J., Boone, D. A., Larsen, J. A., Smith, D. G., and Sangeorzan, B. (1999). Issues of importance reported by persons with lower limb amputations and prostheses. *Journal of Rehabilitation Research and Development*, 36:155–163.

Lerner, A. and Soudry, M. (2011). *Armed conflict injuries to the extremities: A treatment manual*. Springer Science & Business Media.

Lin, S.-J., Winston, K. D., Mitchell, J., Girlinghouse, J., and Crochet, K. (2014). Physical activity, functional capacity, and step variability during walking in people with lower-limb amputation. *Gait & posture*, 40:140–144.

Lindhe (2014). Lindhe Xtend foot. November, 2014. URL: http://www.lindhextend.com.

Major, M. J., Fatone, S., and Roth, E. J. (2013). Validity and reliability of the berg balance scale for community-dwelling persons with lower-limb amputation. *Archives of Physical Medicine and Rehabilitation*, 94:2194–2202.

Major, M. J., Twiste, M., , Kenney, L. P. J., and Howard, D. (2011). Amputee independent prosthesis properties–A new model for description and measurement. *Journal of Biomechanics*, 44:2572–2575.

McGeer, T. (1990). Passive dynamic walking. 9:62–82.

McGowan, C. P., Kram, R., and Neptune, R. R. (2009). Modulation of leg muscle function in response to altered demand for body support and forward propulsion during walking. *JBiomech*, 42(7):850–856.

Miller, W. C., Deathe, A. B., Speechley, M., and Koval, J. (2001a). The influence of falling, fear of falling and balance confidence on prosthetic mobility and social activity among individuals with a lower extremity amputation. *Archives of Physical Medicine and Rehabilitation*, 82(9):1238–1244.

Miller, W. C., Speechley, M., and Deathe, A. B. (2002). Balance confidence among people with lower limb amputation. *Physical Therapy*, 82:856–865.

Miller, W. C., Speechley, M., and Deathe, B. (2001b). The prevalence and risk factors of falling and fear of falling among lower extremity amputees. *Archives of Physical Medicine and Rehabilitation*, 82(8):1031–7.

Moe-Nilssen, R. and Helbostad, J. L. (2005). Interstride trunk acceleration variability but not step width variability can differentiate between fit and frail older adults. *Gait & Posture*, 21(2):164–70.

Moore, S. P. and Marteniuk, R. G. (1986). Kinematic and Electromyographic Changes That Occur as a Function of Learning a Time-constrained Aiming Task. *Journal of Motor Behavior*.

Muir, S. W., Berg, K., Chesworth, B., Klar, N., and Speechley, M. (2010a). Balance impairment as a risk factor for falls in community-dwelling older adults who are high functioning: a prospective study. Technical report.

Muir, S. W., Berg, K., Chesworth, B., Klar, N., and Speechley, M. (2010b). Quantifying the magnitude of risk for balance impairment on falls in community-dwelling older adults: A systematic review and meta-analysis. *Journal of Clinical Epidemiology*, 63:389–406.

Nagai, K., Yamada, M., Uemura, K., Yamada, Y., Ichihashi, N., and Tsuboyama, T. (2011). Differences in muscle coactivation during postural control between healthy older and young adults. *Archives of Gerontology and Geriatrics*, 53:338–343.

Nagano, H., Begg, R., and Sparrow, W. A. (2013). Ageing effects on medio-lateral balance during walking with increased and decreased step width. *EMBC*, pages 7467–7470.

Nordin, E., Moe-Nilssen, R., Ramnemark, A., and Lundin-Olsson, L. (2010). Changes in step-width during dual-task walking predicts falls. *Gait & Posture*, 32(1):92–97.

Novacheck, T. F. (1998). The biomechanics of running. *Gait & Posture*, 7(1):77–95.

O'Connor, S. M. and Kuo, A. D. (2009). Direction-dependent control of balance during walking and standing. *Journal of Neurophysiology*, 102(3):1411–1419.

O'Connor, S. M., Xu, H. Z., and Kuo, A. D. (2012). Energetic cost of walking with increased step variability. *Gait & Posture*, 36(1):102–107.

Ortega, J. D., Fehlman, L. A., and Farley, C. T. (2008). Effects of aging and arm swing on the metabolic cost of stability in human walking. *Journal of Biomechanics*, 41(16):3303–3308.

Össur. Vari-Flex. September, 2014. URL: http:// www.ossur.com/prosthetic-solutions/products/feet/ feet/ vari-flex.

Panzenbeck, J. T. and Klute, G. K. (2012). A powered inverting and everting prosthetic foot for balance assistance in lower limb amputees. 24(4):175–180.

Paysant, J., Beyaert, C., Dati, A.-M., Martinet, N., and Andr, J.-M. (2006). Influence of terrain on metabolic and temporal gait characteristics of unilateral transtibial amputees. *Journal of Rehabilitation Research and Development*, 43(2):153.

Portnoy, S., Kristal, A., Gefen, A., and Siev-Ner, I. (2012). Outdoor dynamic subject-specific evaluation of internal stresses in the residual limb: Hydraulic energy-stored prosthetic foot compared to conventional energy-stored prosthetic feet. *GP*, 35(1):121–125.

Pratt, G. and Williamson, M. (1995). Series elastic actuators. In *Proceedings of the IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS)*.

Pratt, J., Carff, J., Drakunov, S., and Goswami, A. (2006). Capture point: A step toward humanoid push recovery. In *Proceedings of the IEEE-RAS International Conference on Humanoid Robots (Humanoids)*, pages 200–207.

Pratt, J., Krupp, B., and Morse, C. (2002). Series elastic actuators for high fidelity force control. *Industrial Robot: An International Journal*, 29:234–241.

Quesada, R. E., Caputo, J. M., and Collins, S. H. (2015). Increasing ankle push-off work with a powered prosthesis does not necessarily reduce metabolic rate for transtibial amputees. in review.

Ramstrand, N. and Nilsson, K.-A. (2009). A comparison of foot placement strategies of transtibial amputees and able-bodied subjects during stair ambulation. *Prosthetics and orthotics international*, 33:348–355.

Risley, T. R. (1970). Behavior modification: An experimental-therapeutic endeavor. *Behavior Modification and Ideal Mental Health Services*, pages 103–127.

Roaas, A. and Andersson, G. B. J. (1982). Normal range of motion of the hip, knee and ankle joints in male subjects, 30–40 years of age. *Acta Orthopaedica Scandinavica*, 53:205–208.

Roy, A., Krebs, H. I., Williams, D. J., Bever, C. T., Forrester, L. W., Macko, R. M., and Hogan, N. (2009). Robot-aided neurorehabilitation: a novel robot for ankle rehabilitation. *IEEE Transactions on Robotics*, 25:569–582.

Ruina, A., Bertram, J. E. A., and Srinivasan, M. (2005). A collision model of the energetic cost of support work qualitatively explains leg sequencing in walking and galloping, pseudo-elastic leg behavior in running and the walk-to-run transition. 237:170–192.

Sala, A. and Esparza, A. (2003). Reduced-order controller design via iterative identification and control. *European Journal of Control*, 9(1):105–117.

Schmitz, A., Silder, A., Heiderscheit, B., Mahoney, J., and Thelen, D. G. (2009). Differences in lower-extremity muscular activation during walking between healthy older and young adults. *Journal of Electromyography and Kinesiology*, 19(6):1085–1091.

Segal, A. D. and Klute, G. K. (2014). Lower-limb amputee recovery response to an imposed error in mediolateral foot placement. 47(12):2911–2918.

Segal, A. D., Zelik, K. E., Klute, G. K., Morgenroth, D. C., Hahn, M. E., Orendurff, M. S., Adamczyk, P. G., Collins, S. H., Kuo, A. D., and Czerniecki, J. M. (2012). The effects of a controlled energy storage and return prototype prosthetic foot on transtibial amputee ambulation. *Human Movement Science*, 31:918–931.

Selinger, J. C., O'Connor, S. M., Wong, J. D., and Donelan, J. M. (2015). Humans can continuously optimize energetic cost during walking. *Current Biology*, 25(18):2452–2456.

Shultz, A. H., Mitchell, J. E., Truex, D., Lawson, B. E., and Goldfarb, M. (2013). Preliminary evaluation of a walking controller for a powered ankle prosthesis. In *Proceedings of the IEEE International Conference on Robotics and Automation (ICRA)*, pages 4823–4828.

Shumway-Cook, A., Ciol, M., Hoffman, J., Dudgeon, B., Yorston, K., and Chan, L. (2009). Falls in the medicare population: incidence, associated factors, and impact on health care. *Physical Therapy*, 89(4):1–9.

Song, S., LaMontagna, C., Collins, S. H., and Geyer, H. (2013). The effect of foot compliance encoded in the windlass mechanism on the energetics of human walking. pages 3179–3182.

Stinus, H. (2000). Biomechanics and evaluation of the microprocessor- controlled c-leg exoprosthesis knee joint. *Zeitschrift fur Orthopadie und Ihre Grenzgebiete*, 138(3):278–282.

Sup, F., Varol, H. A., Mitchell, J., Withrow, T. J., and Goldfarb, M. (2009). Self-contained powered knee and ankle prosthesis: Initial evaluation on a transfemoral amputee. In *Proceedings of the IEEE International Conference on Rehabilitation Robotics (ICORR)*, pages 638–644.

Tedrake, R., Manchester, I. R., Tobenkin, M., and Roberts, J. W. (2010). Lqr-trees: Feedback motion planning via sums-of-squares verification. *International Journal of Robotics Research*, 29(8):1038–1052.

Tideiksaar, R., Feiner, C. F., and Maby, J. (1993). Falls prevention: the efficacy of a bed alarm system in an acute-care setting. *MSJM*, 60(6):522–527.

Towns, J., Cockerill, T., Dahan, M., Foster, I., Gaither, K., Grimshaw, A., Hazlewood, V., Lathrop, S., Lifka, D., Peterson, G. D., Roskies, R., Scott, J. R., and Wilkins-Diehr, N. Xsede: Accelerating scientific discovery.

Vallery, H., Veneman, J., van Asseldonk, E., Ekkelenkamp, R., Buss, M., and van Der Kooij, H. (2008). Compliant actuation of rehabilitation robots. *IEEE Robotics & Automation Magazine*, 15(3):60–69.

van der Krogt, M. M., Bregman, D. J. J., Wisse, M., Doorenbosch, C. A. M., Harlaar, J., and Collins, S. H. (2010). How crouch gait can dynamically induce stiff-knee gait. *Annals of Biomedical Engineering*, 38:1593–1606.

Van Hedel, H. J. a. and Dietz, V. (2004). The influence of age on learning a locomotor task. *Clinical Neurophysiology*, 115:2134–2143.

van Schooten, K. S., Sloot, L. H., Bruijn, S. M., Kingma, H., Meijer, O. G., Pijnappels, M., and Van Dieën, J. H. (2011). Sensitivity of trunk variability and stability measures to balance impairments induced by galvanic vestibular stimulation during gait. *Gait & Posture*, 33:656–660.

Vegter, R. J. K., Lamoth, C. J., De Groot, S., Veeger, D. H. E. J., and Van Der Woude, L. H. V. (2014). Inter-individual differences in the initial 80 minutes of motor learning of handrim wheelchair propulsion. *PLoS ONE*, 9(2):21–24.

Vittas, D., Larsen, T., and Jansen, E. (1986). Body sway in below-knee amputees. *ProsOrthInt*, 10(3):139–141.

Voloshina, A. S., Kuo, A. D., Daley, M. A., and Ferris, D. P. (2013). Biomechanics and energetics of walking on uneven terrain. *Journal of Experimental Biology*, 216:3963–3970.

Vuillerme, N. and Boisgontier, M. (2008). Muscle fatigue degrades force sense at the ankle joint. *Gait & Posture*, 28:521–524.

Wang, J. M., Fleet, D. J., and Hertzmann, A. (2009). Optimizing walking controllers. *ACM Transactions on Graphics (SIGGRAPH)*, page 1.

Warwick, K. (1996). *An Introduction to Control Systems*. World Scientific, Singapore, 2nd edition.

Waters, R. L. and Mulroy, S. (1999). The energy expenditure of normal and pathologic gait. *Gait & Posture*, 9:207–231.

Wentink, E. C., Prinsen, E. C., Rietman, J. S., and Veltink, P. H. (2013). Comparison of muscle activity patterns of transfemoral amputees and control subjects during walking. *Journal of neuroengineering and rehabilitation*, 10:87.

White, O. R. (1972). *A manual for the calculation and use of the median slope: A technique of progress estimation and prediction in the single case*. Eugene, OR: University of Oregon, Regional Resource Center for Handicapped Children.

Whittle, M. (1996). *Gait Analysis: An Introduction*. Butterworth-Heinemann, Oxford, United Kingdom.

Winter, D. A. (1990). *Biomechanics and Motor Control of Human Movement.* John Wiley & Sons, Inc., Toronto, Canada, 2nd edition.

Winter, D. A. (1991). *The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological.* Waterloo Biomechanics, Waterloo, Canada.

Wisse, M., Schwab, A. L., van der Linde, R. Q., and van der Helm, F. C. T. (2005). How to keep from falling forward; elementary swing leg action for passive dynamic walkers. *IEEE Transactions on Robotics*, 21:393–401.

Witte, K. A., Zhang, J., and Jackson, R. W. (2015). Design of two lightweight torque-controlled ankle exoskeletons. In *Proceedings of the IEEE International Conference on Robotics and Automation (ICRA)*, page accepted.

Zarrugh, M. Y., Todd, F. N., and Ralston, H. J. (1974). Optimization of energy expenditure during level walking. *European Journal of Applied Physiology and Occupational Physiology*, 33:293–306.

Zelik, K. E., Collins, S. H., Adamczyk, P. G., Segal, A. D., Klute, G. K., Morgenroth, D. C., Hahn, M. E., Orendurff, M. S., Czerniecki, J. M., and Kuo, A. D. (2011). Systematic variation of prosthetic foot parameter affects center-of-mass mechanics and metabolic cost during walking. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, 19:411–419.

Zhang, C., Zhu, Y., Fan, J., Zhao, J., and Yu, H. (2015). Design of a quasi-passive 3 dofs ankle-foot wearable rehabilitation orthosis. *Bio-Medical Materials and Engineering*, 26:647–654.

Zidarov, D., Swaine, B., and Gauthier-Gagnon, C. (2009). Quality of life of persons with lower-limb amputation during rehabilitation and at 3-month follow-up. *Archives of Physical Medicine and Rehabilitation*, 90(4):634–645.

Ziegler-Graham, K., MacKenzie, E. J., Ephraim, P. L., Travison, T. G., and Brookmeyer, R. (2008). Estimating the prevalence of limb loss in the united states: 2005 to 2050. *Archives of Physical Medicine and Rehabilitation*, 89:422–429.

Zurada, J. M. (1994). Sensitivity analysis for minimization of input data dimension for feedforward neural network. *Circuits and Systems*, 6:447–450.