34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

Preliminary Study of the Effect of User Intent Recognition Errors on Volitional Control of Powered Lower Limb Prostheses

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Abstract- **Previously developed user-intent-recognition (UIR) systems have demonstrated promising accuracy for identifying the user's locomotion mode, which is potentially useful for volitional control of powered artificial legs in ambulation. The fundamental question facing us now is whether or not the prosthesis users are safe when the UIR system is directly integrated with the intrinsic controller to operate powered artificial legs. In this preliminary study, we aimed to address this question by investigating the effect of UIR errors on the walking stability of users, wearing a UIR-controlled powered transfemoral (TF) prosthesis. First, a novel control of powered prosthesis was developed, which hierarchically connected our designed UIR system with an intrinsic controller. Three types of errors were purposely added into the UIR output at different gait phase while an able-bodied subject walked on a treadmill with the powered prosthesis. Subjective opinions were obtained to evaluate the effect of applied UIR errors on the user's walking balance. The kinematics and kinetics of the prosthetic knee were quantified while the errors occurred. The preliminary results showed that not all the UIR errors applied caused a subjective feeling of balance instability. The effects of UIR errors on the prosthesis control and user's balance depended on the gait phase when the errors happened and the amount of mechanical work applied to the knee joint caused by the errors. The results of this study could aid the future design of true bionic prostheses that enable lower limb amputees to perform various activities intuitively and safely.**

I. INTRODUCTION

RECENT advance in powered lower limb prostheses [1-4] has enabled lower limb amputees to easily and efficiently perform a variety of activities, such as stair climbing, that are difficult tasks for amputees wearing passive devices to perform [5]. Usually, the intrinsic control of powered lower limb prostheses is based on finite-state impedance control [3, 5-6], in which the artificial ankle/knee joint impedance is manipulated based on the gait phase for locomotion tasks or movement state for non-locomotion tasks. These gait phases or movement states were modeled as the states of a finite-state machine (FSM).

Since the desired joint impedance also depends on the type of tasks performed, a user-machine interface is required to switch the intrinsic control parameters based on the user's intended tasks. Manual interface has been available, but is cumbersome to use. Several recent studies [2, 7-10] have reported various user-intent-recognition (UIR) systems for automatic identification of user intent for powered lower limb prosthesis control. These studies all reported 90% or higher accuracy for recognizing the user's intent for switching locomotion modes [2, 7, 10], performing tasks such as sit-to-stand and stand-to-sit transitions [2], and moving artificial joints [8-9].

However, errors in the UIR system were still observed. Will these UIR errors cause the erroneous operation of prostheses? Will the incorrect prosthesis control cause the users to stumble or even fall in the weight-bearing situation? Altogether, the fundamental question is that whether or not the prosthesis users are safe when the UIR system is directly integrated with the intrinsic controller to operate powered artificial legs. To the best of our knowledge, a very limited number of studies have reported the effects of UIR errors on the performance of lower limb prosthesis and user's safety. Varol *et al.* observed that a UIR error that falsely classified the standing mode as the walking mode did not affect the prosthesis control and performance [2]. However, only one type of error was reported, and no further investigation was provided.

In order to address this fundamental question, in this preliminary study, we investigated three types of UIR errors at different gait phases and with varied durations on the performance of a powered TF prosthesis and the stability of user's walking balance. The results of this study could aid the future design of powered TF prosthesis and improve the safety and reliability of prosthesis control.

II. METHODS

A. Design and Control of a Powered TF Prosthesis

A prototype of TF prosthesis with a powered knee joint was designed in our group [11] (Fig. 1). The knee joint was constructed by a moment arm and a pylon and was driven by a DC motor (RE 40, Maxon, Switzerland) through a ball screw. The mechanical sensors on the prosthesis were used for intrinsic prosthesis control. The knee joint angle and angular velocity were measured by a potentiometer instrumented on the knee joint and an encoder connected to the motor, respectively. The 6-DOF load cell (PY6, Bertec, OH) mounted on the prosthetic pylon measured the ground reaction force. The designed powered prosthesis was tethered and controlled by a desktop PC. All the sensor measurements were sampled at 100Hz by a multi-functional DAQ card (PCI-6259, National Instruments, TX). The DAQ board also provided a D/A for control output to drive the DC motor through a motor controller (ADS50/10, Maxon, Switzerland).

The architecture of prosthesis control consists of two levels: a high-level UIR system and a low-level FSM

Manuscript received March 29, 2012. This work was partly supported by NSF #1149385, DoD/TATRC #W81XWH-09-2-0020, NIH #RHD064968A, and NSF #0931820.

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TABLE I. LIST OF STUDIED ERRORS

States	STF			STE			PSW			SWF			SWE		
Error Types (W-)	ST[1]	RA[2]	RD[3]	ST[1]	RA[2]	RD[3]	ST[1]	RA[2]	RD[3]	ST[1]	RA[2]	RD[3]	ST[1]	RA[2]	RD[3]

Note: ST, RA, and RD represented standing, ramp ascent, and ramp descent, respectively.

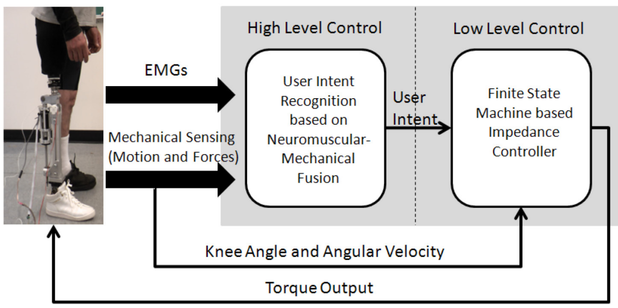

Fig. 1. The architecture of powered prosthesis control

impedance controller [3, 5-6] (i.e. intrinsic controller) (Fig. 1). The UIR system interprets the user's intended task that further modulates the desired joint impedance in the FSM impedance control. In this preliminary study, the considered tasks included level-walking, ramp ascent, ramp descent, and standing. For the first three locomotion tasks, the FSM impedance controller consists of five states (gait phases): stance flexion (STF), stance extension (STE), pre-swing (PSW), swing flexion (SWF), and swing extension (SWE). For the task of standing, FSM impedance controller includes two states: weight bearing and non-weight bearing. The transition between states in FSM was triggered by the ground reaction force and knee joint position [3, 6]. In each state, desired prosthesis joint impedance was defined to mimic the knee impedance in healthy subjects. To match the desired joint impedance, the appropriate joint torques τ_i were calculated based on (1) and generated by the motor.

$$\tau_i = k_i(\theta - \theta_{ki}) + b_i\dot{\theta} \tag{1}$$

where k_i , b_i ,and θ_{ki} denoted the linear stiffness, damping coefficient, and equilibrium point for the i^{th} state, which were the desired joint impedance parameters. θ and $\dot{\theta}$ represented measured knee angle and angular velocity, respectively.

B. Participants and Experimental Protocols

This study was approved by Institutional Review Board (IRB) of the University of Rhode Island and with informed consent of the subject. One male able-bodied subject, free from orthopedic or neurological pathologies, was recruited. A special adaptor was made so that the subject can wear the prosthesis. The subject had received 30 hours of treadmill walking training with the powered prosthesis prior to the experiment.

The desired joint impedance parameters for each studied tasks (including level/treadmill walking, ramp ascent, ramp descent, and standing) were calibrated for the subject before the experiment. A UIR simulator programmed in LabView was designed to simulate the function of the high-level prosthesis control described in Section IIA (also see Fig. 1). This program was capable of producing intended task mode to modulate the desired joint impedance in the low-level intrinsic controller and generating the erroneous UIR decisions at pre-defined gait phases with pre-defined durations. In this study, we considered three types of UIR errors that mistakenly identified the treadmill walking (W) as (1) standing (W-ST), (2) ramp ascent (W-RA), and (3) ramp descent (W-RD), respectively. These three error types were applied to five gait phases (states) in the treadmill walking mode (Table I). Furthermore, the effect of different error duration (100ms, 200ms, and 300ms) was also investigated.

In each trial, the powered prosthesis was first preset in the level/treadmill walking mode. The subject was asked to walk on a treadmill (ActiveStep Simbex, Lebanon, NH) at a self-selected speed with the powered prosthesis. A harness system and parallel bars were used to protect the subject from falls. Then, the UIR simulator generated decision errors at different gait phase and with certain durations to adjust the prosthesis impedance control. In each trial, one type of error with a specific duration was applied to one gait phase. At least ten errors were simulated in one trial. The number of gait cycles between two applied errors was a random integer within the range from 3 to 5. Rest periods were allowed between trials to avoid fatigue. Besides the trials with UIR errors, trials without any error were also conducted to provide a baseline. The trial sequence was randomized. The knee angle and angular velocity, and the torque output at the knee joint were measured by the sensors on the powered prosthesis. All the measurements were sampled at 100Hz and synchronized.

C. Evaluation of Error Effects

The effects of UIR errors on the user's walking balance were evaluated subjectively. After errors were introduced, the subject was asked to give a qualitative assessment of stability by reporting him feeling stable or unstable. In this study, any UIR error that caused the feeling of unstable balance in the subject was defined as the *critical errors*, which should be avoided in the UIR system for the prosthesis control.

The effects of UIR errors on the prosthetic knee control were quantified by the kinematics and kinetics of prosthetic knee. The amount of mechanical work change [12] at the knee joint caused by the UIR errors was quantified. The mechanical work was calculated as the time integral of the knee joint torque multiplied by the joint angular velocity [13]. Additionally, the knee angle in the gait cycle when the errors happened was compared to the knee angle in the gait cycle without the UIR errors.

III. RESULTS

Table II shows the subjective report of walking stability when different types of UIR errors with different durations were applied to the studied phases. Interestingly, the subject reported "stable" when the applied UIR errors had 100ms

TABLE II. SUBJECTIVE RESULTS

States		STF			STE			PSW			SWF			SWE		
Error Types		1	2	3	1	2	3	1	2	3	1	2	3	1	2	3
Durations (ms)	100															
	200	*	*	*				*			*					
	300	*	*	*		*		*			*					

Note: "*" indicates when the "unstable" was reported. 1, 2, and 3 denote the three error types (W-ST, W-RA, and W-RD) respectively (see Table I).

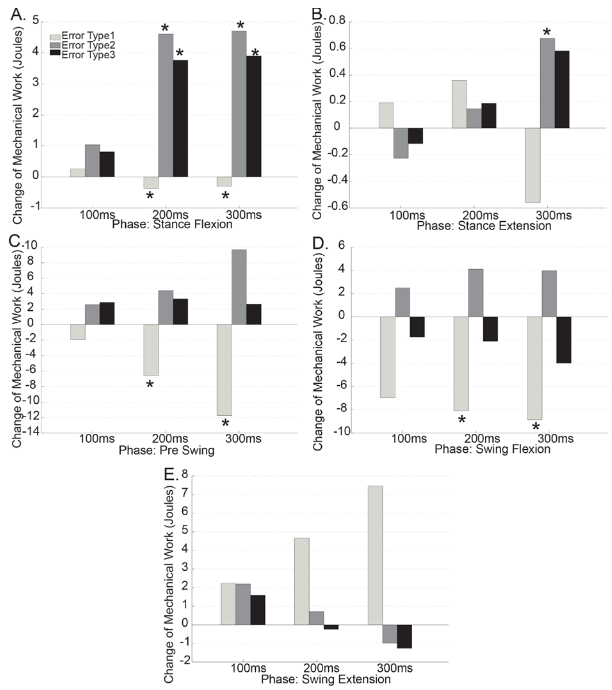

Fig. 2. The amount of mechanical work change caused by different types of errors in different phases and with varied error durations. "*" indicates a critical error reported in Table II.

durations or the errors were located at swing extension phase. When the error duration reached 200ms or more, the reported *critical errors* included all the errors in stance flexion phase, UIR errors that misrecognized walking as ramp ascent in stance extension phase, and errors that misrecognized walking as standing in both pre-swing and swing flexion phases.

Fig. 2 showed the change of mechanical work at the knee joint caused by different UIR errors. Generally, the *critical errors* that elicited the insecurity of balance in the subject had large influence on the net power generated by the powered prosthesis, so larger change of mechanical work was observed in these cases. Moreover, it is noteworthy that the feeling of unstable balance was also pertinent to the gait phase when the UIR errors were applied. For example, in the stance flexion phase (as shown in Fig.2A), errors that caused -0.3 J change of mechanical work were reported as the *critical errors*, while in the pre-swing phase (shown in Fig. 2C), the UIR errors that caused about over -2 J change of mechanical work did not cause the subject to feel unstable.

Fig. 3 showed four examples of comparison of normalized knee joint angle with and without UIR errors. The black lines in Fig. 3 were the knee angle averaged across all the applied errors with the same type, applied gait phase, and duration.

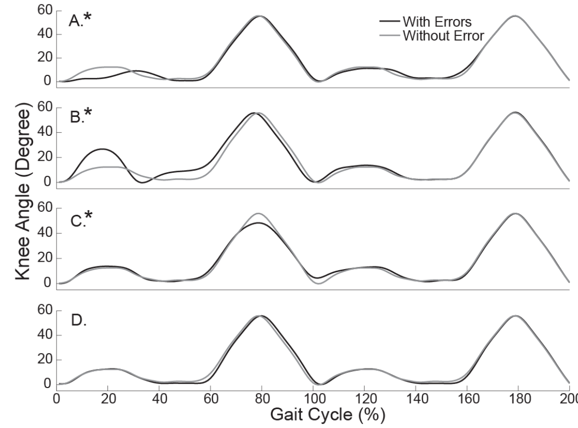

Fig. 3. Comparison of normalized knee angle over two gait cycles with and without UIR errors. A: level-walking misclassified as standing during STF; B: level-walking misclassified as ramp ascent during STF; C: level-walking misclassified as standing in SWF; D: level-walking misclassified as ramp descent in SWE. Error duration=300ms. "*" indicates a critical error reported in Table II.

The knee angle in the gait cycle when the errors occurred and the following cycle was presented. The gray lines were the average knee angle recorded when no UIR error was applied. When walking was misclassified as standing in stance flexion phase (as shown in Fig. 3A), insufficient and delayed knee flexion in the stance phase was observed; when the walking was misrecognized as ramp ascent in stance flexion phase (Fig.3B), excessive stance flexion was generated. If level/treadmill walking was misidentified as standing in the swing flexion phase (Fig. 3C), insufficient knee swing flexion was observed, compared to that of normal gait cycles without applying UIR errors. In Fig. 3D, the UIR errors did not affect the knee angle profile. Clearly, the UIR errors that were reported as the *critical errors* by the subject consistently affected the knee angle profile, compared to the knee angle in the gait cycles without UIR errors. Meanwhile, the errors that did not elicit the user's feeling of imbalance also did not demonstrate significant change of the knee kinematics.

IV. DISCUSSION

The evaluation of the effect of UIR errors on the stability of walking balance was obtained from the subject's feedback. Not all of the UIR errors caused the insecurity of balance in the prosthesis user. This result implies purely using the accuracy or error rate for recognizing user intent is insufficient to demonstrate the potential of the UIR system for volitional control of powered prostheses. Instead, we suggested the identification of the *critical errors* that elicit the subjective feeling of balance instability to quantify the UIR system performance. Although some *critical errors* may not directly cause the user to fall, they will lower the user's confidence in using the powered device; therefore, engineering efforts should focus on targeting and eliminating these *critical errors* to optimize the UIR system design.

It was observed that the effects of UIR errors on the user's feeling of balance and performance of prosthetic knee depended on the phases when the errors happened and also the change of mechanical work caused by the errors. It was reported that humans have different level of demand for balance among different gait phase, which is related with "phase-dependent modulation of reflexes" [14-15]. In the stance flexion phase, the prosthesis was required to generate stance flexion to provide cushioning at heel strike and reduce the rise of the body's center of mass to allow for more efficient gait [16]. However, if too much mechanical work was caused by errors (e.g. errors that misclassified walking as ramp ascent or descent in Fig.2A), it would result in the buckling of the prosthesis (e.g. Fig. 3B) in the stance, which caused the user to feel unsafe. Meanwhile, if insufficient work was caused by errors (e.g. errors that misclassified walking as standing in Fig. 2A), the knee demonstrated an insufficient knee flexion in stance phase, which might cause the sudden rise of user's center of mass [17]. During the pre-swing and swing flexion phases, the stability of the subject may be disturbed when the desired positive work was not generated (shown in Fig. 2C and D), which led to insufficient knee flexion in swing phase (Fig. 3C).

The results of this study might shift the paradigm for evaluating the UIR system and aid the future design of volitionally-controlled powered artificial legs. However, this study is preliminary; several limitations were identified. First, only one able-bodied subject was recruited. It is necessary to investigate the study question on lower limb amputees. Second, this study only considered the condition that errors happened during the treadmill walking mode. The investigation of more types of errors should be included in the future. Third, the full-body kinematics should be monitored in future studies to precisely evaluate the subject's walking stability.

V. Conclusion

The effects of the UIR errors on the volitional control of powered lower limb prostheses were investigated in this study. A prototype of powered TF prosthesis was used as a laboratory test bed. Three different types of errors at different gait phases and with varied durations were simulated when an able-bodied subject wearing a powered prosthesis walked on the treadmill. The preliminary results showed that not all the UIR errors caused the feeling of unstable walking balance in the prosthesis user. The effects of UIR errors for the prosthesis control depended on the phases when the errors happened and the amount of mechanical work applied to the knee caused by the UIR errors. However, this study was preliminarily conducted on an able-bodied subject. In our next study phase, we will investigate other types of UIR errors, quantify their effects in locomotion mode transitions, and test our new prosthesis control on lower limb amputees.

Acknowledgment

The authors thank Ding Wang and Lin Du at the University of Rhode Island, Michael Nunnery and Becky Blaine at the Nunnery Orthotic and Prosthetic Technology, LLC, for their suggestion and assistance in this study.

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