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Toward the Development of a Neural Interface for Lower Limb Prosthesis Control

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***Abstract*— Lower limb amputees form a large portion of the amputee population; however, current lower limb prostheses do not meet the needs of patients with high-level amputations who need to perform multi-joint coordinated movements. A critical missing element is an intuitive neural interface from which user intent can be determined. Surface EMG has been used as control source for upper limb prostheses for many years; for lower limb activities, however, the EMG is non-stationary and a new control strategy is required. This paper describes the work completed to date in developing a novel lower limb neural interface.**

I. Introduction

Commercially available lower limb prostheses can be divided into three categories — mechanically passive devices, microprocessor-controlled passive devices, and powered devices. The majority of these devices are mechanically passive. The movement of passive prosthetic joints relies on the effects of ground reaction forces and on the properties of the mechanical components, such as hydraulic valves, pneumatic valves, or sliding joints. Users must make extra movements with their trunk, pelvis, and residual limb to control the prosthesis. Such control significantly limits the functions of the prosthesis—especially for those with amputations at the knee or higher.

Microprocessor-controlled passive transfemoral prostheses employ sensors and a microcomputer to modulate the resistance of the knee joint [1-4]. The controller receives kinematic and kinetic information from the prosthesis, detects the gait phase, and adjusts the resistance of the knee joint through a hydraulic damper [5] or by modifying a magnetic field [6]. The desired joint resistance is predetermined in each gait phase based on normal gait studies and adjustments made with the prosthetist. A finite state-based control mechanism is used to ensure safe prosthesis control [1, 7, 8]. Studies show that compared to the conventional passive prosthesis, a computerized prosthesis with varied knee joint resistance allows reduced energy consumption, improved smoothness of gait, and decreased hip work production during locomotion [9-11]. In addition, the computerized prosthesis can be programmed for several modes of joint resistance, such as a walking mode, joint locking mode, descending stairs mode and/or free joint mode. Therefore, the user can more easily perform other activities beyond walking by choosing the appropriate impedance mode. For all passive knee prostheses, the joints dissipate energy but cannot provide any net power during normal locomotion modes such as stair climbing. The inability of transfemoral prostheses to deliver joint power significantly impairs the ability of these prostheses to restore many locomotive functions, including ascending stairs and slopes and walking backwards, all of which require significant net positive power at the knee joint, ankle joint, or both [12-18]. Furthermore, even during level walking, transfemoral amputees exhibit asymmetric gait kinematics, expend approximately 60% more metabolic energy relative to healthy subjects [19], and exert as much as three times the affected-side hip power and torque relative to healthy subjects [12], which results in significantly increased socket interface forces. These limitations have a direct impact on the function and quality of life of many lower limb amputees, and most likely contribute to the development of degenerative musculoskeletal conditions.

The function of lower limb prostheses could be greatly improved by the implementation of powered prosthetic knees and ankles, driven by active actuators [20-22]. Powered prostheses can generate joint torque and could allow more efficient performance of activities such as ascending stairs. Current control of powered motion is mode-based. For example, in the POWER KNEE™ [20], the controller drives the powered knee joint differently in level ground walking mode than in stair ascent mode. For each control mode, the control system operates the knee joint based on a predetermined kinematic profile, the motion of the sound leg in the previous step as indicated by an instrumented foot orthosis, and the current kinematics and kinetics of the prosthesis. Hence, in order to switch tasks, the user must "tell" the prosthesis the intended control mode. However, voluntary control is limited; no neural control is available. For example, switching from level walking to stair mode requires the transfemoral amputee to stop and rock back and forth on the prosthesis; this is not intuitive and

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sometimes causes falls when unwanted modes are accidentally activated.

Electromyographic (EMG) signals are a rich source of neural information and have been used as a control source for powered upper limb prostheses for many years. In general, a user voluntarily generates a sustained muscular contraction, control information (i.e. the amplitude, or rate of change of the amplitude) is extracted, and then the appropriate degree of freedom is actuated. Targeted muscle reinnervation (TMR) is a recently developed surgical technique which has been performed successfully in over 30 upper limb patients to provide additional EMG control information [23, 24]. During TMR surgery, residual nerves from an amputated limb are transferred to alternative muscles that are no longer biomechanically functional. The alternative muscles are denervated, and the residual nerves eventually grow into them. The reinnervated muscles then serve as biological amplifiers of neural commands sent through the residual nerves. Thus, there is now a mechanism available for restoring lost control sites in a natural and intuitive manner.

EMG has not been used as a control source for lower limb prostheses because 1) powered, self contained, lower limb prostheses are only now becoming available, and 2) the cyclic nature of EMG signals during gait is highly non-stationary and cannot be directly used with existing myoelectric control schemes. This work describes work completed to date in developing surface EMG neural interfaces for powered lower limb prostheses.

II. CONTROL FRAMEWORK

The control of the device is divided into two layers 1) a neural control mode, and 2) an intrinsic control mode.

A. Neural Mode Control

As mentioned previously, the cyclic nature of gait yields highly non-stationary EMG patterns; however, for very short time durations, the EMG signals may be considered quasi-stationary. Huang made use of this idea to create a gait phase dependent locomotion mode classifier [25]. Heel-contact and toe-off were the gait events used as trigger points to create four phase-dependent classifiers (pre and post toe-off and pre and post heel-contract as shown in Figure 1a). The classifiers were built based on four time-domain EMG features and the linear discriminant analysis (LDA) algorithm. When EMG signals were recorded from muscles controlling the hip, knee, and ankle of able-bodied subjects, the average classification error in four windows was within 10% (black bars in Figure 1b). To simulate a long transfemoral amputation, EMG recordings from below the knee on the able-bodied subjects were removed (gray bars in 1b). The resultant classification error increased 2-3%. Interestingly, the classification errors derived from two subjects with long transfemoral amputations were similar to those derived using the above-knee muscles in able-bodied subjects (Figure 1b).

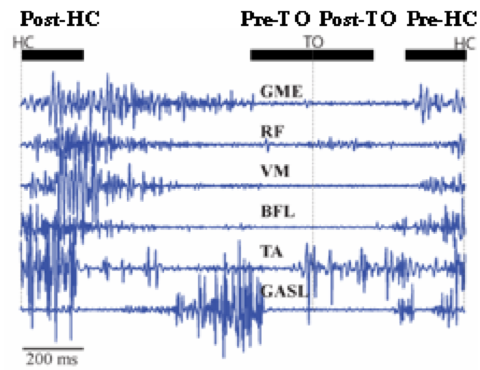

Figure 1: (a) Raw EMG in one gait cycle. Four phase windows aligned with heel-contact (HC) and toe-off (TO) were selected. (b)

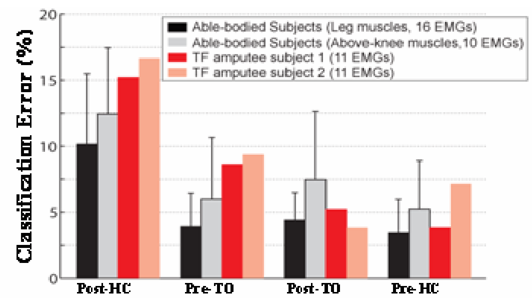

Figure 1: (b) Classification error for identifying seven task modes derived 8 able-bodies subjects (black and gray bars) and two subjects with long transfemoral amputations.

These results demonstrated that the phase-dependent approach to pattern recognition design is feasible for intuitive mode selection in lower limb prostheses. Furthermore, from Figure 1b, it is evident that 16 channels of surface EMG perform better than 10 channels (above-knee muscles) of surface EMG for the able-bodied subjects. These higher classification accuracies suggest that TMR could add additional control information for above-knee amputees. Cadaver studies are currently underway to optimize surgical techniques for above-knee TMR.

B. Intrinsic Control

The Intrinsic Control system provides basic control of cyclical activities once the mode has been selected. The intrinsic control system is based on work by Sup [26], and generates joint torques rather than joint angles. This finite state based impedance control scheme characterizes knee and ankle behavior as a series of finite states consisting of passive spring and damper behaviors. The energy is delivered to the user by switching between appropriate equilibrium positions (of the virtual springs) in each finite state. By doing so, the prosthesis is guaranteed to be passive within each state, and generates power by switching between states. Since the state changes are triggered by the user, the result is a predictable controller that will always default to passive behavior.

III. Current Work

Data is currently being collected with a custom built instrumented passive limb (Figure 2). This prosthesis has a custom socket using vacuum suspension with a goniometer for hip angle and velocity readings. The leg has a Mauch S&S (Manufactured by OSSUR) knee unit instrumented with a goniometer for knee position and velocity readings. The leg uses a six degrees-of-freedom load cell to acquire loading responses between the ankle and knee. Furthermore, multiple EMG electrodes are be attached to the socket and all signals are acquired synchronously with the physical sensor data. The leg can be connected to a variety of feet. Pressure loading on the foot is measured by an insole instrumented with force-sensitive resistors (FSRs). This platform allows us to automatically determine gait phase and integrate EMG signals from transfemoral amputees with on-board sensor data.

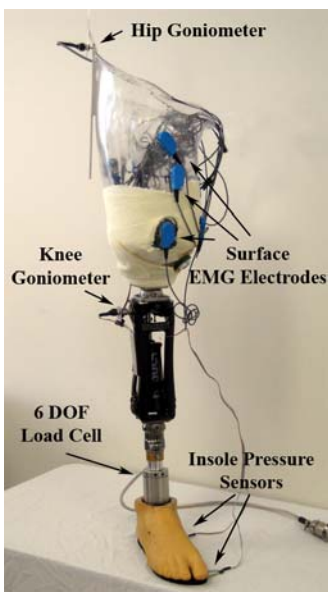

Figure 2: First prototype of RIC experimental passive prosthesis.

Conclusion

EMG controlled upper limb prostheses have been used for many decades; however, to date there has not been robust neural interface for passive or powered lower limb prostheses. This work highlights some recent results which showed that a phase based classifier can be used to predict lower limb gait activities with high accuracies. This work will ultimately lead to a safe and robust neural interface for powered lower limb prostheses.

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