A Biomechatronic EPP upper-limb prosthesis teleoperation system implementation using Bluetooth Low Energy

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***Abstract*— In this paper a real time, stand-alone wireless Biomechatronic Extended Physiological Proprioception (EPP) teleoperation system was implemented using two Bluetooth Low Energy (BLE) wireless Systems on Chip (SoCs). This system is designed to achieve kinesthetic coupling between the amputee and prosthetic arm without the use of the classic EPP mechanical linkage, but with the use of a wireless implementation of a Master/Slave teleoperation topology. The experimental real-time implementation achieved a high level of transparency with minuscule time delays.**

I. Introduction

Control topologies of upper limb prosthesis are focused in improving the life of patients by substituting the functionality of the natural limbs with solutions aesthetically acceptable and risk free for the health of the patients. It is due to these criteria that Extended Prosthesis Proprioception (EPP), a control topology for upper limp prosthesis superior to its rivals in terms of subconscious control, has become abandoned [1]. EPP offers control close to the functionality of the natural limb by allowing the patient to 'feel' the prosthesis in a proprioceptive manner [2, 3]. This is achieved by linking mechanically, using cables, the remaining tendons or muscles of the amputated limb with the prosthesis [4]. In this way, the patient can control the prosthesis without the optical feedback that is needed for other control topologies, such as myoelectric control [4]. Unfortunately, the mechanical linkage is also the greatest disadvantage of this topology. It can be achieved only through surgery, and the result is both aesthetically unacceptable and substantially risky for the patients' health, if cineplasty is used [5].

Implantable components for myoelectric prosthesis control have been proposed in the past using the underlying technology of BIONs [6-9]; however, these proposals lack the inherent advantages of the subconscious feedback found in the EPP topology.

The control topology implemented in this paper is based on the Biomechatronic EPP controller design [5] [10], see Figure 1. The Biomechatronic EPP aspires to eliminate the disadvantages of the traditional EPP controller, without sacrificing the proprioceptive feedback and therefore, the resulting intuitive and superior control. This is achieved by replacing the mechanical linkages (Bowden cables) with a wireless one. We have studied the power and thermal feasibility of the proposed Biomechatronic EPP prosthesis controller in the past, with encouraging results [10]. Using hardware in the loop modules, we have started making real-time performance comparisons of the Biomechatronic EPP versus the Classic EPP and the myoelectric control topologies, with encouraging results for the Biomechatronic EPP topology.

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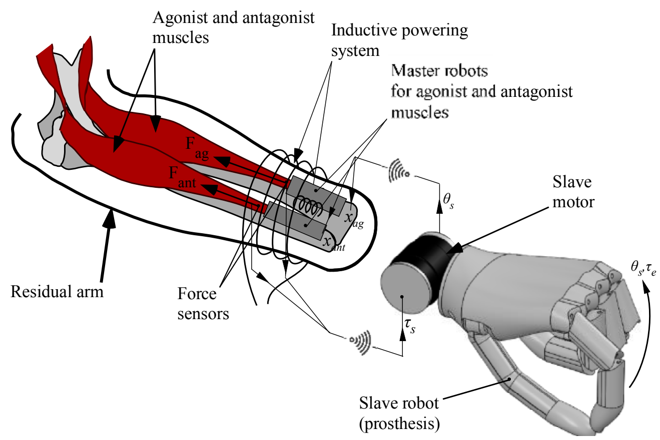

Figure 1. A Biomechatronic prosthetic hand controlled by a wireless Master-Slave teleoperation system [5].

This paper presents research steps along this direction. The objective is to develop an actual stand-alone wireless implementation of the proposed Biomechatronic EPP Controller, resulting in a hand stand-alone real-time portable actual prototype of the proposed controller.

II. Methods

A. Teleoperation System Architectures

The wireless Biomechatronic EPP control topology consists of two Systems on Chip (SoCs) that interact in a Master-Slave bidirectional communication model, using the Bluetooth Low Energy (BLE) protocol. The *experimental setup* built is designed to emulate a *1-DoF prosthesis* shown in Figure 1.

In more detail, the experimental setup, from the Master's side, consists of a *force sensor pair* that provides signals for the forces F_{ag}, F_{ant} that the patient musculotendons would exercise on the 1-DoF prosthesis (Figure 1). With the experimental setup, an operator uses the flexion and

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extension of his wrist joint to provide the input signal to the force sensors. The force input received from the sensors is transmitted from the Master SoC through BLE to the Slave SoC as a voltage driving the Slave DC motor, which emulates the 1-DOF prosthesis (Figure 1, Figure 2).

The displacement of the Slave DC motor is measured and transmitted by the Slave SoC to the Master SoC via the BLE. In turn, the Master SoC controls the Master DC motor to follow the Slave motor displacement. The master DC motor is connected to a leadscrew that translates the rotational to linear displacement, thus providing the operator with feedback by flexing or extending the joint. The implementation of a position controller ensures that the displacement applied as feedback to the operator is proportional to the displacement of the prosthesis. The control loop and the signals in our teleoperation position-force control topology can be seen in Figure 2.

In the bidirectional scenario presented in Figure 1, two Master motors are needed to interface to the agonist and antagonist musculotendon complexes respectively.

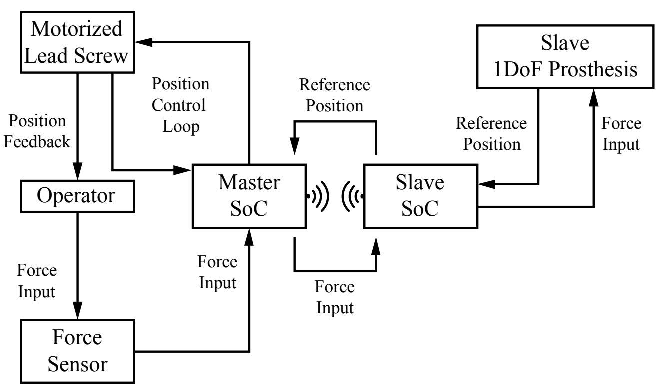

Figure 2. Architecture diagram of the real-time implementation of the Biomechatronic EPP Prosthesis Controller using BLE wireless technology.

B. Hardware Design

In order for the proposed real-time implementation of the Biomechatronic EPP Controller to be realized in the future and used by patients, its Master components should be implantable into the human body. Therefore, the wireless technology to be chosen should not cause problems when interacting with the human tissue. To achieve teleoperation transparency, low latency and high information throughput are needed. The power consumption and the efficiency of the protocol must also be taken into consideration. Bluetooth Low Energy, the prominent technology for wearables, operating at the $2.4GHz$ band was chosen as the most prevalent wireless protocol among other options based on the criteria described in Table I.

The transmit power, the receiving sensitivity, and the power consumption in these operations are the major factors of selection between BLE chipsets. Ease of development was also taken into consideration. The Nordic NRF51822 wireless BLE SoC was chosen for both the Master and Slave components. Based on criteria demonstrated in Table II, the NRF51822 is a good choice compared to other chipsets and has a big advantage that comes as the smallest development supported by the MBED development platform and Arduino. board in the market (RedBearLab BLE nano) and is also

TABLE I. COMPARISON BETWEEN WIRELESS PROTOCOLS.

Wireless Protocols	Efficiency (μW/bit)	Range (m)	Throughput (Kbps)	Latency (ms)
BLE	0.154	280	305	2.5
ZigBee	185.9	100	100	20
Wi-Fi	0.00525	150	6Mbps	1.5
RF4CE	185.9	100	100	20
Ant	0.71	30	29	~0
IrDA	11.7	0.05	1Gbps	25

TABLE II. BLE SOC COMARISON.

SoCs	Tx/Rx (dBm)	CPU (MHz)	Flash/RAM (KB)	RX/TX current (mA) at 0 dBm
Nordic Semiconduct or nRF51822	+4/-93	Core M0 16	256/32	9.7/8
Texas Instruments CC2640	+5/-97	Core M3 48	128/20	5.9/6.1
Cypress PSoC 4 BLE	+3/-92	Core M0 48	256/32	15.6/16.4
Dialog DA14581	+4/-94	Core M0 16	32 OTP /42+8	4.9/4.9

An external microcontroller was used for the control of the Master DC motors. The Master SoC transmitted the position received by the Slave SoC via UART to the microcontroller. A DRV8833 dual H-bridge current control motor driver was used to drive the motors. These drivers are able to provide bidirectional drive and a current up to 1 Amp per motor. Current control is needed in order to provide the operator with “feeling” feedback. For the Master Motors, a Maxon DCX 12 L 12mm was used.

The Slave component of the teleoperation system consists of the NRF51822 and uses an L293D H-bridge to drive the Slave motor.

C. Software Design

A custom BLE Generic Access Profile (GAPP) service was created for the application. The service included two Generic Attribute Profile (GATT) characteristics. The *write* characteristic was used for the transmission of the Slave position to the Master SoC and the *notify* characteristic was used for the transmission of the force input to the Slave SoC. To achieve the highest protocol transmission rate the data was transmitted in 4-byte buffers.

Asynchronous event handling was used for the BLE protocol event as well as the hardware and timer interrupt events. Based on Figure 2, the event loop starts with the analog read of the force input and the transmission of the data to the Slave SoC. This operation is bound to a timer interrupt and occurs every $7.5ms$ (without run time). The Slave SoC is responsible for handling this asynchronous event to activate the Slave DC motor. The position of the Slave motor is calculated from external hardware interrupts and is transmitted to the Master SoC every $7.5ms$ (without run time). The Master SoC handles the asynchronous event and transmits the data to the microcontroller that is responsible for the Master motor control through UART at 9600bps. The UART operations need approximately $3ms$ to be completed.

D. Controller Design

To achieve transparency, the delay of the bidirectional wireless communication must be minimal. Based on the BLE protocol and for small buffers, the minimum interval to exchange data is $7.5ms$. The Master motor must reach the position of the Slave motor before a new setpoint arrives. If we include the latency from both sides in ideal conditions and the UART delay, we have a new Master position setpoint side every 16ms. Adding the software runtime, the desired closed-loop response settling time t_s is $25ms$ which is smaller than period of 100ms (highest frequency of reaching movements is 10Hz).

The dynamics of each Master motor is described by a 2nd order differential equation with respect to position. The actuator linear displacement h is related to its angular displacement by

$$\theta(t) = 2\pi\, x(t) / h \tag{1}$$

Then, neglecting friction, the motor transfer function $G_p(s)$ is,

$$G_p(s) = \frac{X(s)}{I(s)} = \frac{h \cdot K_t}{2\pi \cdot s(Js + b)} \tag{2}$$

where $X(s)$ is the linear output and $I(s)$ the current input, J the inertia seen by the motor, and b the viscous friction. Choosing $\zeta = 0.707$, and for $t_s = 25ms$, the desired closed loop system natural frequency is given by

$$\omega_n = \frac{4}{\zeta t_s} = \frac{5,658}{t_s} = 226 rad / s \tag{3}$$

The desired closed loop poles are then:

$$p_{1,2} = -\zeta\omega_n \pm j\omega_n\sqrt{1-\zeta^2} = -160 \pm j160\, rad / s \tag{4}$$

Given that we implement the controller in an embedded system, for ease of implementation, a PD-controller was designed using a proportional K_P and a derivative K_D gain. The transfer function of the controller is given by:

$$G_C(s) = K_P + K_D s \tag{5}$$

The gains K_P and K_D of the controller are calculated by placing the poles of the characteristic equation

$$1 + G_C(s)G_p(s) = 0 \tag{6}$$

to match the desired ones, given by (4). Due to the discrete nature of the application, the controller (5) is implemented as

$$\text{Output}=\text{K}_P err[KT] + \text{K}_D\left(\frac{err[KT]-err[(K-1)T]}{T}\right) \tag{7}$$

where $T = 1ms$ is the sampling period of the Master microcontroller, and $err[KT]$ denotes the error at the KT time, $err[(K-1)T]$ denotes the error at $(K-1)T$ time, with $K = 1,2,3...$ increasing at each sampling period by 1. The output of the controller is the PWM duty cycle, which feeds the H-bridge driving the DC Master motor. The error is calculated as the difference between the desired and actual position between the iterations of the control loop.

III. Results

The voltage response of the Slave SoC along with the corresponding Force Input command from the Master SoC is illustrated in Figure 3. The PWM signal has been filtered with a low pass equiripple filter. The delay between these signals is $35ms$. The two signals seem to be highly correlated, which means that the output voltage that drives the Slave (endpoint prosthesis) adequately follows the force command at the Master (muscle).

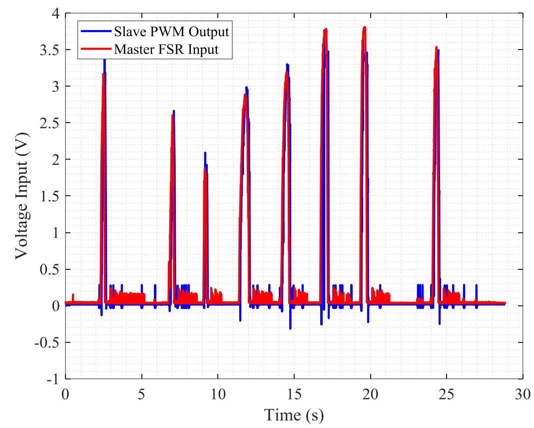

Figure 3. Force input vs Slave Motor Activation.

The position responses of the Master motors are compared with the position command from the Slave SoC, see Figure 4. These responses adequately follow the position command of the Slave Motor. The Master motors start with a delay of $44ms$ to the Slave motor.

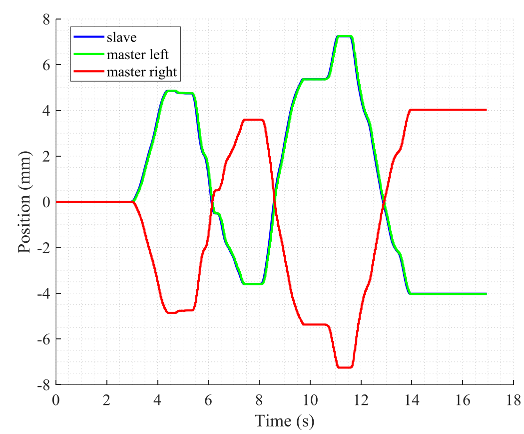

Figure 4. Master motors positions follow Slave motor position response.

More precisely, Figure 5 shows that the position error between the Master and Slave motors is imperceptible.

Based on Figures 3-5, there is adequate evidence pointing to a high degree of transparency in the proposed Master/Slave teleoperation system, as recommended in [11].

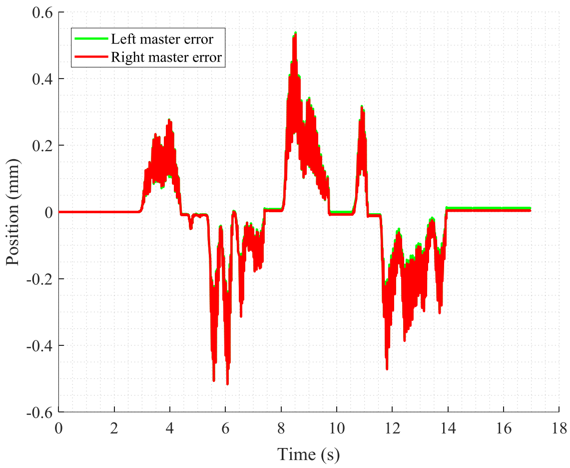

Figure 5. Position error between Master motors and Slave motor.

IV. Discussion

The performance characteristics of the wireless Biomechatronic EPP topology are shown in Table III. A tiny position steady state error was observed. The cumulative latencies at startup and settling time - including software execution, interrupts and callback delays - are in the order of $40-50ms$. The low clock ($16MHz$) and the discrete nature of the controller are responsible for this delay, which can be improved with a faster clock or microprocessor. Nevertheless, the $50ms$ latency is small enough for prosthetic and tracking applications, since brain-to-hand delays has been measured to be of the order of $300-500ms$.

TABLE III. Wireless Biomechatronic EPP Performance Characteristics

Parameter	Wireless Biomechatronic EPP
Position Steady State Error	0.1°
Delay between force input to Slave motor activation	35ms
Delay in settling time between Master and Slave motors	49ms
Percentage max overshoot	0.14%

In this work, we designed and developed a prototype wireless Biomechatronic EPP embedded teleoperation system. We evaluated the performance of the wireless teleoperation system based on the transparency of it and we also compared it to the results of the Biomechatronic EPP, which is a topology superior to the Classic EPP. From the results conducted we proved the transparency of the wireless Biomechatronic EPP controller. Since we developed the actual embedded hardware, then feasibility experiments as far as it concerns low heat, low power and small signal attenuation in human models, can be performed. Further development is needed to create a prototype that can be miniaturized and implantable into the human body.

V. Conclusion

The proposed Biomechatronic EPP topology was implemented using the modern wireless BLE technology. Very small delays and high degree of transparency shown are encouraging to continue work on this controller topology, since it has the inherent value of subconscious feedback of the EPP prosthesis control. The proposed controller can be the building block of a multi-degree of freedom Biomechatronic EPP controller for subconscious control of upper-limb prostheses. We plan to use the controller we built, and characterize its performance as far as signal attenuation, thermal, and power performance are concerned in a similar to the projected implantable in human scenario.

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