Design, development, and evaluation of upper and lower limb orthoses with intelligent control for rehabilitation

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Abstract
World’s ageing population, prevalence of chronic diseases, and shortage of healthcare resources have increased interest in home-based physical therapy and rehabilitation. However, the standard equipment is often too bulky and unsuited for home use. This paper presents the design, development, and evaluation of a set of wearable orthoses for home-based physical therapy. Originatively utilising miniature electromagnetic brakes, the system can deliver a low to moderate range of resistive torques suitable for isotonic, isometric, and open-chain resistance exercises. In human tests with the orthoses, low-level muscle activation and low- to moderate-level energy expenditure were achieved. This suggests the system’s potential for use in a wide range of applications, including postoperative treatments for muscle injury and the early stages of rehabilitation, and tele-rehabilitation. The orthoses were integrated with a video game console to form a new research platform that can be used to study the effects of force haptics on muscle activation and energy expenditure in exergames. The data obtained aid the future development of exergame prescription standards.

1 | INTRODUCTION

The global population is ageing rapidly because of a declining birth rate and increased life expectancy. This unprecedented and prevalent demographic shift is exerting a heavy burden on healthcare systems worldwide. For instance, in 2017, 962 million people globally were aged 60 or older; this figure is projected to reach 1.4 billion in 2030 and 2.1 billion in 2050 [1]—where, in 2050, the number of older adults will exceed that of children for the first time in human history. Furthermore, the elderly population is itself ageing. Because the incidence of non-communicable disease (NCD) increases with age, the worldwide trend of ageing population has increased NCD prevalence, especially in developed countries. In 2016, an estimated 41 million deaths were caused by NCDs, accounting for 71% of 57 million deaths worldwide [2]. NCDs have become the world’s number one killer as well as the leading cause of death among the elderly population. The most common NCDs are cardiovascular diseases (CVDs), diabetes, chronic obstructive pulmonary disease (COPD), kidney diseases, cancer, Alzheimer’s disease, Parkinson’s disease, and other dementias. Additionally, mental diseases are often comorbid with NCDs. In 2016, CVDs alone killed 17.7 million people, accounting for the largest number of deaths in both the NCD and elderly populations [3, 4]. Stroke is the second leading cause of death and the third leading cause of disability. Each year, 15 million people suffer from stroke [5]. 70% of strokes, and 87% of stroke-related deaths as well as disability-adjusted life years occur in low- and middle-income countries [6]. The number of people with disabilities is also increasing with the ageing population. It is estimated that 15% of the world’s population, approximately 1 billion people, live with one or more disabling conditions. Over 46% of older adults have a disability, and more than 250 million people experience moderate to severe disability [7]. NCD is becoming one of the world’s greatest economic challenges owing to rising healthcare costs, and productivity loss from illness,
disability, and death. By 2030, the global economic costs of CVDs, cancer, diabetes, and COPD are projected to reach US$1 trillion, US$458 billion, US$745 billion, and US$4.8 trillion, respectively. Future economic and human development will eventually be compromised which will have deleterious consequences. Over the next two decades, NCD-related economic loss is expected to reach US$47 trillion, accounting for 75% of the global gross domestic product (GDP) [8].

In response to the escalating demand for chronic care, governments worldwide are devising new approaches to allocating healthcare resources. In this respect, rehabilitation services—such as physical therapy, exercise therapy, and occupational therapy—all play a vital role. Excluding acute and remitting conditions, and those associated with mild disability, approximately 75% of the total number of years living with a disability (YLDs) are linked to health conditions for which rehabilitation is beneficial [9]. Despite efforts to accelerate the supply of services through community-based rehabilitation, the demand for rehabilitation services continues to exceed availability, resulting in a large unmet need. According to the World Federation of Occupational Therapists, the recommended minimum number of occupational therapists per million population is 750. However, in 71 countries, the number of registered occupational professionals is only one-tenth of this, even in high-income countries [10]. Statistics in the UN Database on the Living Arrangements of Older Persons indicate a current trend where countries [10].

Considering such a context, various groups have developed portable systems in the form of active orthoses to support older adults and those with disabilities in their homes. Some active orthoses function as assistive devices to reduce the forces of gravity on the body during movement, whereas others deliver mechanical resistance for physical training. Unlike fixed weights and resistance bands, active orthoses can be programmed to exert dynamically changing mechanical resistances, and to provide personalised levels of training. They can be easily reconfigured and become force haptic-enabled without change of hardware components. This capability allows much broader choices of applications, higher degree of interaction, and better user experience. These orthoses also provide quantitative data to caregivers as a reference for prescribing subsequent trainings. However, many of these systems are too heavy and bulky due to the fact that the resistance was generated by pneumatic actuators, fluid-based dampers or DC motors. Actuator weight, transmission efficiency, and power supply continue to be the principal technical bottlenecks, and off-the-shelf components often fail to meet the requirements of low weight with high efficiency [13]. Quantitative data on the performance and effectiveness of these active orthosis devices, as well as their health benefits, are lacking. Thus, an analysis of energy expenditure (EE) and muscle activity would aid the objective evaluation and comparison of these systems.

This paper presents a home-based system that was designed and developed in our laboratory to provide rehabilitation training and physical exercise for older adults in a home setting. The system is an intelligent and active upper- and lower-limb orthosis system that can be programmed to provide different modes of resistance training while simultaneously monitoring muscle activities and energy expenditure. Different from the orthoses developed by other groups, our system utilises miniature electromagnetic brakes (EBs) rather than pneumatic devices to deliver resistance, and it enables the simultaneous measurement of EE and muscle activation. Despite the EBs’ small size and low power consumption, they can deliver the adequate resistive force for a range of rehabilitation applications. The system can be expanded into an experimental exergaming system for further research. Section 2 presents the design, development, and resistive torque measurements. The system was tested with human participants to evaluate its functional suitability for use in rehabilitation and exercise training for older adults. In Sections 3 and 4, the experimental protocol, data handling, and signal processing techniques are described. Section 5 presents the experimental results. Sections 6 and 7 then discuss the performance of the system and its possible development in various applications.

2 | SYSTEM DESIGN, DEVELOPMENT, AND EVALUATION

2.1 | System overview

Figure 1 displays the setting and building blocks of the system. The data acquisition (DAQ) subsystem comprises the following: a metabolic system, which captures EE signals; a biopotential amplifier, which acquires surface electromyography (sEMG) measurements from selected muscles; and an angle recorder, which captures the angles of the elbow and knee joint during limb movements. Through a control subsystem driven by a power subsystem, the orthoses can be programmed to exert varying levels of mechanical resistance for customised training. The PC-based control terminal is connected to the data.
acquisition subsystem and control subsystem via USB interfaces. The terminal coordinates and synchronises system operations such as starting and stopping data acquisition, displaying and storing the received signals, and sending control commands to the orthoses.

2.2 | Upper- and lower-limb orthoses

As depicted in Figure 2(a), the upper-limb orthosis allows one degree of freedom (DOF) of movement about the elbow. Parts of the orthosis are adjustable to better fit the arm of an adult user. The principal mechanical components are aluminium alloy trestles and stainless-steel assemblies. The length of the orthosis from elbow to wrist is adjustable and can extend from 30 to 40 cm at the forearm and from 30 to 40 cm from the shoulder to elbow at the upper arm. The inner diameter is adjustable from 15 to 35 cm at the forearm and from 20 to 40 cm at the upper arm. Equipped with interchangeable shafts and mounts, the elbow joint is designed for compatibility and for easy attachment by three models of EB—namely, MBGS09AA, MBG082AA, and MBG084AA (Chain Tail, Taiwan). These EBs vary in size and provide different ranges of passive resistive torque at the elbow joint, allowing full range and 1-DOF flexion and extension of the upper limb.

Unlike the active orthoses developed by other groups, our system uses EBs in place of motors or pneumatic devices. An EB operates by generating a magnetic field when an excitation voltage is applied, drawing the rotor plate against the stator friction surface. Instead of exerting forces as motors do, an EB applies resistive forces passively. Compared with DC motors, EBs generate higher torque at lower power consumption for the same weight and volume ratio. Interfacing with EBs requires only a few electronic components. Some recent lightweight EBs have been used in wearable applications [14]. To prevent slippage of the orthosis relative to the upper limb during abrupt motion, our design incorporated i) a freely sliding and rotational shaft and adjustable lock ring mechanism in the structure of the orthosis, ii) multiple Velcro straps, and iii) a shoulder strap. In addition to the EB, the elbow joint is also integrated with an analogue angular displacement sensor WDD35D4-5K (Lollette, China), which has 5 kΩ resistance and ±15% total resistance tolerance as well as a wiring duct for the interconnection of sensors and actuators. Figure 2(b) presents the lower-limb orthosis, the design of which is similar to that of the upper-limb orthosis. This design allows 1-DOF movement about the knee joint. The same three choices of EB models and angular displacement sensor are mounted at the knee joint.

2.3 | Control subsystem

The control subsystem comprises two Arduino MEGA 2560 microcontroller unit (MCU) development boards; both are connected to an L293D Motor Driver Shield integrated with an H-bridge to drive the EBs. Because the EBs require a nominal voltage of 24 V at a power of 6–10 W, a voltage booster circuit in the power subsystem was used to drive them (and also the control subsystem). Input to the subsystem can be interfaced with the control terminal via USB or Bluetooth, and the output ports of the subsystem directly connect to the EBs. The development boards are pre-programmed to receive the control terminal’s serial-format commands (9600 baud, 8 bits, no parity, 1 stop bit); these commands indicate the resistive torque that needs to be delivered. Upon receiving the commands, the subsystem generates two streams of pulse width modulation (PWM) signals to...
drive the EBs. A higher duty cycle entails a stronger resistive torque generated by the EB.

2.4 Data acquisition subsystem

The data acquisition subsystem comprised (i) a biopotential amplifier MP36 (BIOPAC, USA) for acquiring two-channel sEMG signals at a 2000 Hz sampling rate and 24-bit resolution, (ii) a portable metabolic (gas analysis) system K4b² (COSMED, Italy) to capture the EE signal, and (iii) an Arduino UNO microcontroller (A/D channel) to record the angular displacement at the elbow and knee joints at a 10 Hz sampling rate and 10-bit resolution. The K4b² system comprised a flexible face mask for sampling the participant’s expired air for the measurement of ventilation, oxygen, and carbon dioxide concentrations. Operations of the aforementioned subsystems were synchronised by a trigger signal generated by the Arduino unit.

2.5 Control terminal

The PC-based control terminal centrally coordinates all operations in the system. The terminal acquires and stores the sEMG, angle, and EE signal data from the data acquisition subsystem while simultaneously sending control commands to the control subsystem to drive the EBs. At the terminal, the user can coordinate the tasks with individual software packages (RS232 terminal, BIOPAC Student Lab, COSMED K4b² software). A LabVIEW program was also developed. The program takes the angle signal as input and outputs control commands to the control subsystem to actuate the EBs and implement real-time force-haptic control. This feature is reserved for advanced applications, such as exergame-based and force-haptic rehabilitation programmes.

2.6 Resistive torque measurement

Prior to human tests, our team investigated the resistive torques of the EB-enabled orthoses. The objectives were to i) identify the ranges of resistive torques that could be delivered by the three target EB models, and ii) identify the EB models that are suited to rehabilitation/physical training for older adults. In initial trials, each EB model was attached to an orthosis frame and was driven by a PWM signal with a duty cycle ranging from 39% to 100%. As shown in Figure 3, one section of the orthosis was fixed to a digital torque meter ZQ-11B HP-100 (Zhiqu Precision Instruments, China) while the other section was rotated about the EB joint. The procedure was repeated 20 times during which the maximum and average torques were recorded.

For a duty cycle <50%, the torque was negligible. The EB models MBGS04AA, MBGS09AA, and MBGS02AA were found to exert resistive torques of 0.10–0.83, 0.07–1.69, and 0.65–11.32 Nm, respectively [15]. Therefore, the range of torques exerted by MBGS09AA and MBGS02AA were too small for our target application. MBG084AA, however, covered the range required for rehabilitation applications and muscle training. For example, assuming that the distance between the elbow and wrist is 30 cm, a torque of 11.32 Nm about the elbow joint corresponds to a 3.8 kg load on the hand. MBG084AA was therefore chosen for the subsequent development of our system.

The subsequent experiment investigated the resistive torques exerted by MBG084AA, which was installed in the upper- and lower-limb orthoses during flexion (decreasing angle) and extension (increasing angle) about the joint. These angles were measured while the unit was driven at an increasing duty cycle (50%–85% at 5% increments). Each measurement was repeated 20 times. Figure 4(a, b) present the relationships between resistive torque and duty cycle; these relationships were for the upper-limb and lower-limb orthoses, respectively. The mean and standard deviation of the resistive torque readings are also presented in the figures. We found that, in both orthoses, the resistive torque increased nonlinearly with the duty cycle; the resistive torque was generally higher during extension than during flexion. The maximum resistive torque exerted by the lower-limb orthosis was 4.3 Nm higher than that exerted by the upper-limb orthosis. These findings were used as reference data in subsequent calibrations of the system.

3 EXPERIMENTAL PROTOCOLS FOR HUMAN PARTICIPANTS

For the initial trials, 23 and 21 healthy male participants were recruited to test the upper-limb and lower-limb orthoses, respectively. The experiments involved 5-min repeated and paced flexions/extensions of the upper limb about the elbow and of the lower limb about the knee, while the EB was driven at an 80% duty cycle. Under a resistive torque >8 Nm, the biceps brachii in the upper limb was activated at >40% of maximum voluntary isometric contraction (MVIC) in 15 participants, and the triceps brachii was activated at >40% MVIC in 17 participants. In trials with the lower-limb orthosis, duty cycles of 70%, 80%, and 90% were attempted, achieving activation of the rectus femoris at up to 83%, 89%, and 92% MVIC, respectively. Although driving the EBs at a high duty cycle resulted in substantial muscle activation, all participants found it difficult to move the orthoses smoothly, and some experienced muscle
Fatigue 1 min after the test began. Because the orthoses were designed for the rehabilitation of older adults and of patients with chronic conditions, it was decided that the EBs should be driven at a lower duty cycle (<70%) to prevent injury and to provide a more realistic range of resistive torques for the target application. Using the reconfigured setup, a set of experiments were conducted with a pool of 15 healthy male participants (mean age of 26.6 years and standard deviation of 8.72 years) to investigate the levels of exercise intensity and muscle activities that could be achieved with the orthosis system. The experiments involved simultaneous measurement of sEMG and EE under different orthosis settings. The experimental results provide insights into the relationship between levels of muscle activation, EE, and the resistive torque of the orthoses. Ultimately, the findings will provide a reference for specific applications and target user groups that will benefit from the system. Details of the experiments are provided in the following section.

3.1 Experiment with upper-limb orthosis

Each participant initially rested for 15 min in a seated position. The preparation of electrode sites involved disinfecting the skin by rubbing it with alcohol. Two sets of Ag–AgCl bipolar electrodes were attached on the upper right arm to capture two channels of sEMG signals, one from the biceps brachii and the other from the triceps brachii. Having had their arm secured to a fixture on a table and having been given consistent verbal encouragement, the participant performed MVIC for 30 s to enable sEMG signals to be recorded from the two sets of muscles. Isometric contraction involves no movement in the joints, although the muscles contract to provide a force. The sEMG signals captured during MVIC were used to normalise sEMG-related indices in subsequent submaximal contractions by the same participant. In the next phase of the investigation, the upper-limb orthosis and the K4b2’s face mask were attached to the participant. EE was measured for 3 min while they were resting. The participant was then instructed to perform repeated flexion (approximately 140°) and extension (0°) about the elbow for 7 min under the guidance of a metronome. The procedure was performed four times, each at a randomly assigned duty cycle of 0%, 55%, 59%, or 63%. These duty cycles corresponded to the resistive torques of approximately 0, 0.9, 4.7, and 7.0 Nm, respectively. Participants rested for 20 min between sessions. The exercise sessions were denoted as T0 (0 Nm) to T3 (7.0 Nm). Figure 5(a) illustrates the experimental setup.

3.2 Experiment with lower-limb orthosis

A similar procedure was employed for the experiment with the lower-limb orthosis. The participant first rested for 15 min in a seated position. Bipolar electrodes were then attached to the right thigh to capture sEMG signals from the rectus femoris. With their right leg tied to a fixture on the chair, the participant performed MVIC in a concentric direction for 60 s while sEMG measurements were taken. The lower-limb orthosis and
K4b² face mask were then attached to the participant, following which EE was measured for 3 min while the participant was at rest. The participant was then instructed to perform repeated 1.8-s flexion (approximately 90°) and 1.8-s extension (0°) about the knee for 7 min while EE and sEMG measurements were taken. The procedure was performed four times, each at a randomly assigned duty cycle of 0%, 55%, 59%, or 63%. These duty cycles corresponded to resistive torques of approximately 0, 6.8, 9.8, and 10.5 Nm, respectively. Participants rested for 20 min between sessions. The exercise sessions were denoted as T4 (0 Nm) to T7 (10.5 Nm). Figure 5(b) illustrates the experimental setup.

4  SIGNAL PROCESSING AND ANALYSIS

All signal processing and analyses were performed using MATLAB. Each sEMG signal sampled at 2000 Hz was processed by a bandpass filter with a passband of 50–250 Hz and a 50 Hz notch filter. Following full-wave rectification, the root mean square (RMS) signal $x_{RMS}(n)$ was produced with a 200 ms window [16]. An example of an sEMG measurement and its corresponding RMS signal is presented in Figure 6. For experiments with the orthoses, sEMG signals captured during 30 consecutive gait cycles within the third and fourth minute of the 7 min session were segmented for analysis. For each of the 54-s segments, an RMS signal, $x_{RMS}(n)$, and an average RMS value, $x_{RMS}_{avg}$, were derived. To analyse and compare the level of muscle activity among the participants, the average RMS values $x_{RMS}_{avg}$ were normalised according to the corresponding mean RMS obtained from participants in previous MVIC sessions. The resulting metrics were expressed in terms of a percentage of MVIC and denoted as %MVIC.

The raw EE signal samples from the K4b² system were expressed in terms of metabolic equivalent tasks (METs). The signal samples were not separated by a fixed time interval; instead, the sampling times were dependent on the varying respiratory rate of the participant. Therefore, each 7-min EE signal was first interpolated and resampled at 1 Hz with an antialiasing lowpass filter. Motion artefact was observed at the beginning and end of the recordings due to the participant moving the face mask; therefore, samples in the first and last 10 s of each EE recording were omitted.

Figure 7 presents examples of the processed EE signals from participants in experiments with the (a) upper-limb orthosis and
TABLE 1  Level of activation (%MVIC) of the biceps brachii and triceps brachii in exercise sessions T0 to T3 and of the rectus femoris in exercise sessions T4 to T7. Mean and maximum %MVIC of the 15 participants were calculated.

| Upper Limb | Level of muscle activation (%MVIC) | Biceps | Triceps |
|------------|-----------------------------------|--------|---------|
| T0         | Mean                              | 6.847  | 4.190   |
|            | Max                               | 16.349 | 13.476  |
| T1         | Mean                              | 6.394  | 4.017   |
|            | Max                               | 13.249 | 10.590  |
| T2         | Mean                              | 7.183  | 5.852   |
|            | Max                               | 16.058 | 10.515  |
| T3         | Mean                              | 8.947  | 7.696   |
|            | Max                               | 12.474 | 12.934  |

| Lower Limb | Rectus Femoris | Mean | Max |
|------------|----------------|------|-----|
| T4         |                 | 5.687| 13.872|
| T5         |                 | 5.696| 10.500|
| T6         |                 | 5.943| 11.143|
| T7         |                 | 7.916| 13.331|

(b) lower-limb orthosis, respectively. To analyse EE trends over time, each EE signal was divided into three 120-s segments, corresponding to the start (0–2 min), midpoint (2–4 min), and end (4–6 min) of the exercise session. The mean and maximum EEs in each segment were then calculated. Because each of the 15 participants underwent eight sessions of exercise, a total of 360 mean EE values and 360 maximum EE values were derived from the data. These values were used to analyse the trends. The EE values were already normalised to body weight because the K4b2 system was calibrated to each participant prior to the experiments.

5 | RESULTS

Table 1 summarises the levels of muscle activation of the 15 participants, which were recorded in the tests with orthoses at different settings (T0 to T7). In experiments with the upper-limb orthosis, the mean %MVIC in the biceps ranged from 6.847% MVIC in Session T0 to 8.947% MVIC in Session T3, whereas the mean %MVIC in the triceps ranged from 4.190% in Session T0 to 7.696% MVIC in Session T3. Under the same resistive torque settings, considerably higher %MVIC was achieved by some participants: maximum %MVIC was 16.349% and 13.476% MVIC in the biceps and triceps, respectively. Figure 8(a) and b present boxplots depicting the distribution of average %MVIC in Sessions T0 to T3. An evident trend of %MVIC increasing with resistive torque was evident for both set of muscles.

In the experiments with the lower-limb orthosis, mean %MVIC in the rectus femoris ranged from 5.687% in Session T4 to 7.916% MVIC in Session T7. The maximum muscle activation ranged from 10.500% to 13.872% MVIC. The boxplot in Figure 9 presents the distribution of values in Sessions T4 to T7. We noted a trend of %MVIC increasing with resistive torque.

Table 2 summarises the mean and maximum values of the average EE of the 15 participants in the upper-limb exercise sessions T0 to T3. From the beginning to end of each session, the mean EE value, expressed in terms of METs, increased from 1.397 to 1.578, from 1.324 to 1.443, from 1.368 to 1.575, and from 1.411 to 1.787 in T0, T1, T2, and T3, respectively. The highest average EE was 1.913, 1.781, 2.072, and 2.035 METs in T0, T1, T2, and T3, respectively. The boxplot in Figure 10 illustrates the distribution of the average EE of the 15 participants. These EE measurements were taken during experiments with the upper-limb orthosis at the start, midpoint, and end of the exercise sessions. We noted an evident overall trend of EE increasing with time and with resistive torque. In addition to average EE, highest EE was also a useful measure, which ranged from 1.429 METs in T0 to 5.754 METs in T3.
Table 3 summarises the mean and maximum values of the average EE in the lower-limb sessions T4 to T7. The mean EE increased from 1.607 to 1.685, from 1.484 to 1.618, from 1.512 to 1.698, and from 1.596 to 1.914 METs in T4, T5, T6, and T7, respectively. The highest average EE was 2.296, 2.288, 2.308, and 2.416 METs in T4, T5, T6, and T7, respectively. The box-plot in Figure 11 illustrates the distribution of the average EE of the 15 participants. These EE measurements were taken during experiments with the lower-limb orthosis. We noted an evident overall trend of EE increasing with time and with resistive torque. Throughout the investigation, the maximum EE ranged from 1.976 in T4 to 3.280 METs in T7.

In summary, with the upper-limb orthosis delivering a resistive torque of up to 7.0 Nm during exercise, the biceps and triceps were activated at between 4.2% and 8.9% MVIC, and that value occasionally surged above 13.0% MVIC. Use of the orthosis resulted in average and maximum EEs of 2.035 METs and 5.754 METs, respectively. Compared with EB being disabled, when the EB was enabled, average bicep and tricep activation occurred at 2.1% and 3.5% MVIC higher, respectively. Additionally, the average EE increased by 0.159 MET. Findings from
the lower-limb orthosis experiments were similar. While delivering a resistive torque up to 10.5 Nm, the rectus femoris was activated at 5.7%-7.9% MVIC, and occasionally >13.3% MVIC. The average and maximum EE values were 1.914 METs and 3.280 METs, respectively. Compared with EB being disabled, when the EB was enabled, average muscle activation and EE increased by 2.2% MVIC and 0.229 METs, respectively. In general, these results indicate whether and how much the orthosis system can provide physical rehabilitation for older adults.

6.1 DISCUSSION

We verified the ability of our orthosis system to exert low to moderate resistive torque. Our system is thus a feasible aid to rehabilitation; specifically, it enables home-based and individualised isometric exercise, isotonic exercise, and resistance training (RT) of the limbs. Isotonic exercises involve movements based on muscle contraction; along with RT, these exercises are often prescribed in therapeutic modalities that target a range of musculoskeletal conditions, including those which are chronic [17]. Conversely, isometric exercises can be utilised in any course of rehabilitation or home-based exercise programme. They are a preferred option for post-injury and early-stage rehabilitation to avoid pain, swelling and insufficient healing. Isometric exercises are often initially prescribed to increase muscle strength until the patient can progress to isotonic exercises and RT. Studies have also demonstrated that submaximal isometric training significantly increases the strength and cross-sectional area of muscles in older adults [18]. Our orthosis system can also serve as a configurable setup for open-chain resistance exercises, which are appropriate for all ages. For older adults, such training usually begins with isometric exercise before progressing to eccentric motions and finally to concentric contraction. The resistance, duration, and/or frequency can be increased within each level. Because the orthosis system can be easily programmed for different levels of varying and fixed mechanical resistance, it is suitable for such a scenario.

Low-intensity muscle activation is defined as that at <20% MVIC [19–20]. The EE ranges for low- and moderate-intensity exercises are 1.1–2.9 METs and 3.0–5.9 METs, respectively [21]. With reference to these benchmarks, the experimental findings presented in Section 5 suggest that our orthosis system can be used as a tool for physical therapies that target low-level muscle activation and involve low- to moderate-intensity exercises. The system has a wide variety of potential applications in training and rehabilitation among older adults. Physical therapy at low %MVIC has been proven to be beneficial in the early and intermediate phases of non-operative and postoperative treatments for patients with muscle disorder and injury [22]. Low-intensity exercise is effective in improving the health of older adults: such exercise encourages compliance, decreases injury risk, and ensures long-term sustainability. Specifically, such exercise improves flexibility, balance, and lower-limb muscle strength, in addition to alleviating depressive symptoms [23]. Emerging research has also suggested that a low- to moderate-level of aerobic activity improves physical function, executive functioning and memory, health-related quality of life, and depressive symptoms in survivors of stroke [24].

Our team is making ongoing efforts to integrate the orthoses into an exergaming platform. Unlike conventional video games that are played with handheld controllers, exergames capture and utilise the movements and gestures of the player’s body as commands, offering a physically active gaming experience [25]. Traditionally, exercises have been the primary interventions for improving physical activity levels to lower the risk of chronic diseases. Recent studies have found that exergames can serve as an alternative physical activity intervention [26]. Because exergames are popular and readily accepted by all ages, scholars generally believe that they have promise in physical therapy and fitness training [27, 28]. In preliminary studies, exergames have been found to help enhance balance control and physical functioning in healthy people, older adults, and patients who had stroke [29, 30].

Despite the potential health benefits, two challenges must be addressed before exergames can be widely deployed in these applications. The first challenge is a lack of force feedback. Most exergaming systems only provide vibration or sensory feedback in response to the motions of the player. Therefore, the level of EE that players can achieve in a gaming session is generally lower than that obtained in a traditional training session [31, 32]. Second, few quantitative studies have investigated the effect of exergame-based physical therapy on bodily functions and health conditions. Although some inter-exergame studies have been conducted, they have only been able to conclude that EE varies greatly between exergame activities [33–36]. Other groups of researchers have studied the relationship between limb movements and EE; however, their results were qualitative, being obtained from general observations. If quantitative benchmarks are absent, health professionals would find it difficult to select appropriate exergames to achieve the targeted health benefits.

Our exergaming platform can therefore serve as a research platform to tackle these challenges. First, it is equipped with force-haptic functionality. The programmable feature of the EBs makes it easy to create a time-varying load in the form of resistive torque in the elbow and knee joints of the orthoses. Depending on the dynamic features of games, the additional load is expected to increase exercise intensity, EE, and the degree of game interaction. To ensure low-cost implementation, we designed our exergaming system to have the orthoses interfacing with a conventional video game console and a PC. During a gaming session, the player wears the orthoses and issues the commands through the flexion and extension of their limbs about the joints.

Signals from the embedded angular displacement sensors are fed into the analogue-to-digital port of an MCU. This MCU continuously and simultaneously performs two tasks: (i) relaying the angular values to the PC via its USB (serial) interface and (ii) issuing commands to the game controller via its I/O pins. As a result, the limb movements are translated into game control signals and the force-haptic feature is realised with a closed-loop feedback control block implemented using LabVIEW. As illustrated in Figure 12, the control block takes the digitised angular
signal from each orthosis as input and then outputs actuator commands to the MCU. In doing so, the control block changes duty cycle of the EB-driving signals to produce time-varying resistances. The characteristics of the haptic response can be modified by changing the transfer function in the LabVIEW control block.

Figure 13 presents the setup for conducting experiments with the exergaming platform. Wearing the orthoses and sensors in a seated position, the participant plays a haptic-enabled car-racing exergame by controlling the car with limb movements. Figure 14 depicts how the MCU serves as an interface between the orthoses and the console controller to translate limb movements into controller key-press signals. To remedy the problem of the lack of quantitative findings (the aforementioned second challenge), our exergaming platform can simultaneously record EE, muscle activity, and limb movement data during gaming sessions. We anticipate that a quantitative study of the effects of force-haptic characteristics on muscle activities and EE will produce benchmark data that can be used to set exergame prescription standards in the future.

**FIGURE 12** Functional blocks of force-haptic operation in the exergaming platform are realised using LabVIEW and MCUs. Closed-loop control takes angle data as input and outputs EB control commands.

**FIGURE 13** Under the guidance of the investigator and wearing the orthoses in a seated position, the participant plays a haptic-enabled car-racing exergame by controlling the car with their limb movements about the joints. Data on muscle activity, EE, and movements are recorded simultaneously.

**FIGURE 14** Signals from the angular displacement sensors are fed into the MCU, which then issues commands via its I/O pins to the game controller to create key-press signals.

### 7 CONCLUSION

In this study, a set of upper- and lower-limb orthoses for home-based physical therapy for the older-adult population were designed, developed, and evaluated. Making use of miniature EBs instead of pneumatic devices, the wearable setup was able to deliver a low to moderate range of resistive torques that are suitable for isotonic, isometric, and open-chain resistance exercises. In experiments with the orthoses involving human participants, low-level muscle activation and low- to moderate-level of EE were achieved. Despite the limited achievable range of exercise intensity and level of muscle activation, the orthoses have the potential to serve in a broad range of applications—including postoperative treatments for muscle injury, exercises for enhancing physical functioning, and early-stage rehabilitation for older adults. The programmable nature of the EBs also enabled the orthoses to perform intelligent control—including providing time-varying loads and force haptics.

Our developed system is a prototype for investigating the technical feasibility and functions. Based on our findings, two future directions have been identified: (i) home-based tele-rehabilitation, in which the orthoses are remotely reconfigured and monitored; and (ii) integration of the intelligent orthoses into an exergaming research platform. To demonstrate the aforementioned benefits of the system, the orthoses were integrated into an exergaming research platform for a quantitative study on the effect of force haptics on muscle activation and EE in exergames. Data obtained using our new research platform will provide a useful benchmark upon which exergame prescription standards can be developed.

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