ICRA 2017 Workshop on Advances and challenges on the development, testing and assessment of assistive and rehabilitation robots: Experiences from engineering and human science research

Abstract

Assistive robots for health and welfare applications are required to display perceptual, cognitive and bodily-kinaesthetic capabilities that are natural and intuitive for older people and persons with disabilities to interact with, communicate with, work with as partners, and learn to adapt to their needs. However, the embodiment of such capabilities has been scarcely studied, so it is still required that the health and social care staff and the user groups could explore and learn how to exploit the capabilities of the assistive robots. Therefore, a multidisciplinary approach to promote the study from the engineering and human science to introduce the next generation of assistive robots is desired. The goal of this workshop is to provide a forum for sharing the experiences from the engineering and human science research on the development, testing and assessment of assistive robots and present the most recent advances and challenges in order to foreseen novel designing approaches and user based studies addressing healthcare and social welfare applications in the ambient assisted living from a world-wide perspective with point of departure in interdisciplinary research collaboration.

The workshop topics include (but are not limited to) the following:

- Rehabilitation Robotics
- Physically Assistive Devices
- Prosthetics and Exoskeletons
- Wearable robotics
- Welfare technology
- Mechatronics systems for assistance and rehabilitation

Invited Speakers

1. Prof. Christine Gustafsson, Mälardalen University College, Sweden
2. Prof. Miki Saijo, Tokyo Institute of Technology, Japan
3. Prof. Jorge Solis, Karlstad University, Sweden
4. Prof Etienne Burdet, Imperial College London, UK
5. Prof. Yukio Takeda, Tokyo Institute of Technology, Japan
6. Prof. Shaoping Bai, Aalborg University, Denmark
7. Prof. Antonio Frisoli, Scuola Superiore Sant'Anna, Italy
8. Prof Lorenzo Masia, Nanyang Technological University, Singapore
9. Prof Herman van der Kooij, Twente University, Netherlands
10. Prof. Eiichiro Tanaka, Waseda University, Japan
11. Prof. Shuichi Fukuda , Keio University, Japan
12. Prof . Kazuyuki Kojima, Saitama University, Japan
13. Prof. Kyoungchul (KC) Kong, Sogang University, Korea
14. Dr. Simona Crea, Scuola Superiore Sant'Anna, Italy

Program

8:30-8:40 WORKSHOP OPENING (Organizers)

SESSION 1: ENGINEERING AND HUMAN SCIENCE RESEARCH TO IMPROVE ROBOTIC SOLUTIONS FOR ASSISTANCE (Chair: Prof. Antonio Frisoli)

8:40-9:00 Prof. Christine Gustafsson, Mälardalen University College (Sweden)
Title: Development and implementation of health and welfare technology in dementia care, the JustoCat experience

9:00-9:20 Prof. Miki Saijo, Tokyo Institute of Technology (Japan)
Title: Evaluation of the usability of AT devices for frail elderly people from the aspect of interaction analysis

9:20-9:40 Prof Etienne Burdet, Imperial College London (UK)
Title: Neuroscience-based, practical rehabilitation of the hand function

9:40-10:00 Prof. Shuichi Fukuda, Keio University (Japan)
Title: Proprioception assistance

10:00-10:30 COFFEE BREAK

SESSION 2(1): NEW ADVANCES IN EXOSKELETONS AND WEARABLE ROBOTIC TECHNOLOGIES FOR ASSISTANCE AND REHABILITATION (Chair: Prof. Eiichiro Tanaka)

10:30-10:50 Prof. Yukio Takeda, Tokyo Institute of Technology (Japan)
Title: Simple and Low-cost Robotic solutions for assistive rehabilitation

10:50-11:10 Prof. Shaoping Bai, Aalborg University (Denmark)
Title: Design and development of exoskeletons for active and assisted living of the elderly

11:10-11:30 Prof. Antonio Frisoli, Scuola Superiore Sant'Anna (Italy)
Title: Design and control of upper limb exoskeleton for rehabilitation and assistance

11:30-11:50 Prof Lorenzo Masia, Nanyang Technological University (Singapore)
Title: Upper limb Soft Wearable Exosuit: Human in the Loop.

11:50-12:10 Prof Herman van der Kooij, University of Twente (Netherlands)
Title: Lower limb exoskeletons and gait rehabilitation

12:10-13:15 LUNCH

SESSION 2(2): NEW ADVANCES IN EXOSKELETONS AND WEARABLE ROBOTIC TECHNOLOGIES FOR ASSISTANCE AND REHABILITATION (Chair: Prof. Jorge Solis)

13:15- 13:35 Prof. Eiichiro Tanaka, Waseda University (Japan)
Title: Development of Life Support Devices Using Inclusive Design

13:35-13:55 Prof. Kyoungchul (KC) Kong, Sogang University (Korea)
Title: Rehabilitation through Entertainment with a Full-Lower Body Exoskeleton Robot, ANGELEGS

13:55-14:15 Dr. Simona Crea, Scuola Superiore Sant'Anna (Italy)
Title: Exoskeleton technologies for upper-limb rehabilitation in stroke

14:15-15:00 Poster teaser + poster session #1 Assistance and Support

1-1
Swagata Das, Yuichi Kurita, and Toshio Tsuji
Design and development of a PAM-enabled wrist assist glove
1-2
Irfan Hussain, Leonardo Meli, Gionata Salvietti and Domenico Prattichizzo
Towards the assessment of the functionality and usefulness of robotic extra-finger
1-3
Ann-Louise Lindborg, Jorge Solis, Miki Saijo, Yukio Takeda, Cheng Zhang, Ryuta Takeda
Design approach of a robotic assistive eating device with a multi-grip and camera for frail elderly's independent life
1-4
Manabu Okui, Shingo Iikawa, Yasuyuki Yamada, and Taro Nakamura
Proposal of Exoskeleton with Variable Viscoelastic Joint System named "Airsist I"
1-5
Shigang Li, Tatsuya Fujiura, and Isao Nakanishi
Gaze-Based Interface for Wheelchair Robot - Acquiring Gaze Trajectory by Full-View Camera
1-6
Susumu Hara
A Networked Control System Using Wi-Fi Communication and Its Application to Assist Cart Control
1-7
Tasuku Hoshino, Miki Yazawa, Hiroshi Miyazawa and Hiroshi Kobayashi
Development of a dynamically stable wheelchair of single spherical drive

15:00-15:15 COFFEE BREAK

15:15-16:00 Poster teaser + poster session #2 Rehabilitation and Exercise

2-1
Daisuke Matsuura,Toshihiro Ichinoseki, Tasuke Ogawa, Yuji Ichikawa, Tomokazu Takakura, Susumu Tachikawa, Koji Yoshihara, Hiroshi Ujiie and Yukio Takeda
Development of Ankle Rehabilitation Device which can Mechanically Avoid Undesirable Load
2-2
Justin Fong, Vincent Crocher, Iven Mareels, Ying Tan and Denny Oetomo
The importance of transparency in rehabilitation robot design and its evaluation
2-3 Jacob Nielsen, Anders Stengaard Sørensen, Thomas Søndergaard Christensen, Thiusius Rajeeth Savarimuthu and Tomas Kulvicius
Individualised and adaptive upper limb rehabilitation with industrial robot using dynamic movement primitives
2-4
Anders Stengaard Sørensen, Jacob Nielsen, Jørgen Maagaard, Jakob L. Nielsen, Gitte Rasmussen, Dennis Day

Natural Kinesthetic Interaction and Social Relations Between Training-Robots and Their Users

2-5

Tadaaki Ikehara

Evaluation of a Device that Combines Exercise and Entertainment -In the view of Electroencephalogram analysis-

2-6

T. Kodama, R. Kasai, Z. Gu, D. Zhang, W. Kong, S. Cosentino, S. Sessa, Y. Kawakami, and A. Takanishi

Stride length estimation in self-pace and brisk walking with a single inertial measurement unit on the shank

2-7

Takumi Ohashi, Makiko Watanabe, and Miki Saijo

An Interaction Analysis of User-Testing to Extract Salient User Experience with the Robotic Assistive Device Life-Walker

SESSION 3: NEW ADVANCES IN MECHATRONICS SYSTEMS FOR ASSISTANCE AND REHABILITATION (Chair: Prof. Yukio Takeda)

16:00-16:20 Prof. Kazuyuki Kojima, Saitama University (Japan)
Title: Floor-Deformation Mechanism for Nursing Bed Considering of Patient's Posture and Position Change

16:20-16:40 Prof. Jorge Solis, Karlstad University (Sweden)
Title: Development of a multipurpose assistive robot vehicle

16:40-17:00 Prof. Jose L. Pons, Spanish National Research Council (Spain)
Title: Effects of associative neurorehabilitation with WRs in stroke and spinal cord injury

17:00 CLOSING OF WORKSHOP

Organizers

- **Jorge Solis, Ph.D. (primary contact person)**
  Associate Professor
  Karlstad University, Sweden
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- **Yukio Takeda, Ph.D.**
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WORKSHOP PROCEEDINGS – PAPERS

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Invited Speakers and Poster Papers

Poster Papers

Development and implementation of health and welfare technology in dementia care, the JustoCat experience

Prof. Christine Gustafsson, Mälardalen University College (Sweden)

We are facing a rapidly expanding market offering welfare technology to health and social care. It is important that users (patients, relatives, professional caregivers) and researchers are active and involved in both development and implementation. The presentation addresses important aspects of this work: Welfare technology corresponding to real needs, Ethical considerations and Scientific evaluations, which are exemplified in a project where researchers developed an interactive robotic cat; JustoCat www.justocat.se

Evaluation of the usability of AT devices for frail elderly people from the aspect of interaction analysis

Prof. Miki Saijo, Tokyo Institute of Technology (Japan)

With the rapid increase in the elderly population, the over-80 market is expected to expand all over the world. In order to foster this emerging market, evaluation of AT devices to support the quality of life of this age cohort is essential, however the methodology for testing the usability of these devices for frail elderly people is not yet established. Since these devices are used in the context of the interaction between frail elderly people and their caregivers, the usability of the devices should be evaluated under real-life conditions with an analysis of this interaction. This presentation will introduce usability tests for a 4-wheel electric assisted bicycle for frail elderly people which were carried out in Kakegawa city Japan in 2015 and the concept behind an ongoing international research collaboration project between Japan and Sweden on “food for life". It is the research project of the development and implementation of robotic assistive devices equipped with multi-grip tools and vision systems designed to help frail elderly people lead independent lives.

Neuroscience-based, practical rehabilitation of the hand function

Prof Etienne Burdet, Imperial College London (UK)

Impressive rehabilitation robots have been developed in the last decades, which had however limited penetration into the clinic, and can hardly be used home. To address the needs of the increasing number of individuals affected by neurological diseases, we propose to develop practical and efficient technological solutions for neurorehabilitation, based on neuroscience investigations. This lecture will present three aspects of this research: i) Sensor-based tools and signal processing techniques to develop evidence-based assessment on the sensorimotor condition; ii) Simple rehabilitation devices for training the hand function that can be used from the bedside to home; iii) Robot-aided investigation of neural mechanisms of motor learning and recovery.

Proprioception assistance

Prof. Shuichi Fukuda, Keio University (Japan)

Coordination of body movements is very important for the handicapped. But motion control is very difficult due to its huge degrees of freedom. In this work, we aimed to help them adjust their sense of balance or proprioception by providing them with motion patterns and with a quantitative measure for evaluating their motions. This will not only facilitate motion learning of the handicapped, but it is expected to motivate them. Thus, this approach is expected to help them to recover not only physically, but also mentally.

Simple and Low-cost Robotic solutions for assistive rehabilitation

Prof. Yukio Takeda, Tokyo Institute of Technology (Japan)

In this talk, the design and development of the wearable walking assist machine using crutches for paraplegic and hemiplegic persons, and ankle rehabilitation device are presented. All of these machines and devices have been designed considering whole system composed of the user's body structure as well as the machine itself. The dof (degree of freedom) of the total system was well considered, and appropriate arrangement of passive joints has been done. Through several experiences of development of devices, current and future directions are discussed for development of useful assistive devices.

Design and development of exoskeletons for active and assisted living of the elderly

Prof. Shaoping Bai, Aalborg University (Denmark)

In this talk, the design and development of the exoskeletons will be presented in connection with an EU project called AXO-SUIT, which is aimed to assist the elderly in their instrumental activities of daily life. This will include the identification of the functional assistance requirements of potential end users of the AXO-SUIT, biomechanical modeling and simulation in human-exoskeleton interaction, the design of lightweight and compact shoulder mechanism, and the development of adaptive control algorithms with novel human intention detection. Initial end-user testing results will be described and discussed.

Design and control of upper limb exoskeleton for rehabilitation and assistance

Prof. Antonio Frisoli, Scuola Superiore Sant'Anna (Italy)

What are the main guidelines and indications to be followed for the design of exoskeleton based rehabilitation systems? In this talk we will face the main construction and implementation solutions, control approaches and we'll show the application in upper limb stroke rehabilitation scenarios and assistance.

Upper limb Soft Wearable Exosuit: Human in the Loop.

Prof Lorenzo Masia, Nanyang Technological University (Singapore)

The presentation will ramp up with an initial summary of my experience in robot aided rehabilitation to gradually shift towards the new trend in robotics which is the wearable assistive exoskeleton. In recent years, compliant actuation technology have been increasingly developed and employed in the fields of robotic rehabilitation, haptics and wearable exoskeletons: devices where safety, limitation of peak forces and gentle interaction are extremely important. To date, several examples of robotic applications have been designed to address the demanding needs of these disciplines that require compliance in actuation and manipulation. In my talk I would like to introduce a new approach in wearable exosuits for upper limb assistance: starting from the concept of a high mechanically transparent device, which allows mobility of the human user, to new architectures based on textiles materials and underactuated compliant transmissions. By introducing new wearable devices for upper limb assistance/augmentation, I will discuss in details our hardware and software solutions with the Human User in The Loop.

Lower limb exoskeletons and gait rehabilitation

Prof Herman van der Kooij, University of Twente (Netherlands)

Several (robotic) devices have been developed to support gait rehabilitation, including lower limb exoskeletons. They differ in which degrees of freedom they control, constrain or leave free. Controlled degrees of freedom differ in how they are controlled and the control performance. Given the large differences in devices, the question can be raised whether these differences do matter. The pro and cons of the different approaches will be discussed from the perspective of human motor control and learning, and from manual physical therapy.

Development of Life Support Devices Using Inclusive Design

Prof. Eiichiro Tanaka, Waseda University (Japan)

We developed various life support devices for the elderly and patients. To design these devices, we used the concept of inclusive design. All devices were designed by not only engineers but also medical doctors, PT, OT, patients, designers, etc. In the case of walking assistance device, we suggest to assist only ankle joint utilizing stretch reflex and the bi-articular muscle, this idea was born from the discussion. In this talk, the method of inclusive design, some developed assistance devices for walking, standing, upper limb operation, and lifting are introduced.

Rehabilitation through Entertainment with a Full-Lower Body Exoskeleton Robot, ANGELEGS

Prof. Kyoungchul (KC) Kong, Sogang University (Korea)

The design of an assistive robot for people with partially impaired walking ability demands unique requirements, such as minimal mechanical impedance and high back-drivability, as well as high power density. Mechanical parts must be ergonomically designed, such that humans can use the devices for a long period of time without discomfort. The motor control function of people with impaired walking ability is not as robust as that of the normal, and thus the overall human body system becomes vulnerable to disturbances (i.e., external forces, inclinations, etc.) or model variations (i.e., loads, etc.). It is, however, still active, unlike complete paraplegics, and thus the robot must not generate any unexpected resistance to the voluntary motions of people with impaired walking ability for the safety and the minimal discomfort. Therefore, both the design and control of assistive robots for people with partially impaired walking ability are very challenging. ANGELEGS is a new wearable robot for assisting the mobility of people with partially impaired walking ability. Assistance as needed (AAN) is a fundamental approach in the control of ANGELEGS. AAN requires the assistive robots to generate assistive joint torques only when the human needs assistance and to exhibit minimal resistance when the assistance is not necessary. For effective rehabilitation treatment, various services including the internet-based rehabilitation program, self-fitness program, and the other entertainment programs are provided with ANGELEGS also.

Exoskeleton technologies for upper-limb rehabilitation in stroke

Dr. Simona Crea, Scuola Superiore Sant'Anna (Italy)

Upper-limb movement disorders due to neurological accidents significantly reduce a patient's quality of life, limiting the independence of the affected subjects. Robotic wearable devices for rehabilitation have been proposed to allow patients to practice longer and more intensively but at the same time, should be designed to provide relatively-high torques with a safe human-robot interface. In this talk the overview of the upper-limb exoskeletons developed at The BioRobotics Institute will be discussed, with a particular focus on the solutions adopted for realizing highly ergonomic devices. Recent results on human-subjects experiments for the validation of the physical and cognitive human-robot interface will be also presented.

Floor-Deformation Mechanism for Nursing Bed Considering of Patient's Posture and Position Change

Prof. Kazuyuki Kojima, Saitama University (Japan)

A nursing bed intending to support positioning a physically handicapped patient is discussed in this presentation. For physically handicapped people such as paraplegia, hemiplegia, and quadriplegia, posture and position change on the bed is one of the most important activities of daily living (ADL). But these activities impose an excessive burden on the patient. In order to reduce such burden, we have been developing the nursing bed to support the patient to change their position and posture on the bed. In this presentation, we show a mechanism of our prototype and experiments for validation of its functionality.

Development of a multipurpose assistive robot vehicle

Prof. Jorge Solis, Karlstad University (Sweden)

In industrialized countries, regional disparities in healthcare and welfare services, increased medical expense caused by aging societies, and shortages of medical staff have become serious problems. Assistive robots are developed to improve the security, independence and quality of the elderly. However, current assistive robots are only designed to accomplish one task and still require technical knowledge from the elderly and/or care givers to exploit the robot's capabilities. Therefore, the next generation of assistive robots has to be natural and intuitive to interact with, communicate with, work with as partners, and learn to adapt to their respective needs. For this purpose, the development of a multipurpose human-friendly assistive robot vehicle has been proposed at Karlstad University. The proposed system is being designed for providing walking-support to the elderly, as well as assistance for carrying-medical tools to care-givers. In this paper, the current development progress of two assistive robot vehicles will be described: an intelligent carrying-medical tools robot vehicle and a human-friendly assistive robot vehicle for supporting physically elderly.

Effects of associative neurorehabilitation with WRs in stroke and spinal cord injury

Prof. Jose L. Pons, Spanish National Research Council (Spain)

Assistive and rehabilitation WRs can lead to improved functional outcomes and subsequent neurophysiological changes when causally driven out of neural motor planning activity. In this talk, the principles of associative interventions with WRs will be outlined and the assessment of functional and neurophysiological changes presented in the context of both upper and lower extremities WRs.

Design and development of a PAM-enabled wrist assist glove

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***Abstract*— Upper limb of the human body has a higher likeliness of getting affected by muscle fatigue and injury, due to repetitive motion. This paper presents an assistive glove to support wrist motions of the human body. The force your hand glove is enabled with pneumatic actuators and stretch sensors which support the user, actively in performing wrist flexion, extension, and pronation. The design of this glove takes the dynamics and behavior of the wrist joint into consideration, for achieving better performance.**

I. Introduction

Wrist joint of the human body is one of the most complex joints and is an important but not indispensable link between the hand and forearm [1]. Nevertheless, the wrist plays an important role in performing day to day activities such as typing, knitting, opening and closing door knobs, etc. While performing heavy duty activities, wrist muscles can be worn out, eventually resulting in pain and numbness. This can also be caused by repetitive activities like, during playing tennis, badminton, and other sports. Moreover, wrist dynamics play a major role in the decisions taken to design any type of rehabilitation product. Apparently, the passive stiffness of the wrist is a major impedance which is to be overcome by the neuromuscular system to rotate the wrist and perform various motions. This paper presents an upgraded version of the force your hand glove [2], which is a novel PAM based wrist assist glove, capable of supporting wrist motions. The proposed model is portable, lightweight, wearable, compatible with wrist dynamics, and above all, active in response to human motion.

II. Related work

Several researchers are considering the development of soft wearable orthosis for the human's upper limb. However, issues such as wearability, portability, and diversity in the degrees of freedom, have not been considered yet. Flexion and extension are the most studied movements despite, supination/pronation and radial/ulnar deviation being equally vital. The soft wearable device proposed in [3] considers rehabilitation of wrist by providing motion specific assistance, but, without synchronization with the user's desired motions. The *RiceWrist* in [4] is an exoskeleton robot to provide training and rehabilitation through kinesthetic feedback in four degrees of freedom. It is comparatively, a heavy system with lesser portability and wearability. A study in [5] shows that the degree of passive stiffness of the wrist joint is anisotropic (different for all directions). The lowest passive stiffness slightly deviates toward radial deviation for wrist extension and toward ulnar deviation for wrist flexion. Therefore, the present literature shows limited work on active rehabilitation of wrist, most of which are heavy exoskeletons, with limited wearability, and consideration of wrist dynamics.

III. Methods

McKibben type of artificial muscles was decided to be used in this work because of their light weight and good performance. The currently developed assist model is enabled with three sets of pneumatic artificial muscles (PAMs) and stretch sensors [6], each of which, supports wrist flexion, extension, and pronation. Fig. 1 gives an overall idea about the placement of PAMs (DAIYA industry corporation), and stretch sensors (YAMAHA corporation).

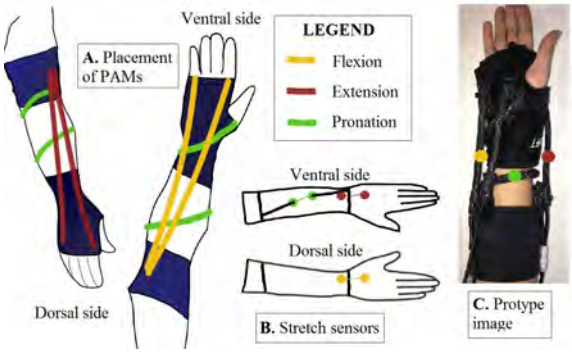

Fig. 1. Force your hand glove design - placement of PAM (pneumatic artificial muscle), stretch sensor, and image of the latest prototype

A. Pneumatic artificial muscles (PAMs)

The artificial muscles used in force your hand glove are comprised of rubber tubes, enclosed in a braided mesh shell. When the muscle is pressurized from one end, it inflates and shortens in length in accordance to the load given at the other end. In this work, experiments were conducted and 0.3MPa was observed as the optimum value of pressure for actuation. The value gave a maximum decrease in length without abnormal leakage from the actuator. The first set of

PAMs for flexion have been placed on the ventral side of the hand. One actuator attaches just below the little finger, and the second near the base of the index finger. The remaining ends are attached at a common point near the ventral side of the elbow. Alignment of PAMs for flexion is done in such a way that their actuation enables the wrist to flex with an ulnar deviation. The two actuators for extension are attached in a similar way on the dorsal side of the hand to empower the wrist in extending towards the radial side. The pronation actuator is spiraled around the forearm, one end of which is near the ventral side of the wrist on the major thumb pad, and the other end attached near the elbow joint.

B. Stretch sensors

Stretch sensors are used to get information about the user's intent at any point in time.Three of them are used to sense each motion. The flexion related stretch sensor is attached on the dorsal side of the wrist towards the radial direction. Extension associated sensor is attached on the ventral side of the wrist towards the ulnar direction. For pronation, after much study, the sensor was attached to an elastic attached from the thumb pad to the elbow.

C. Mechanism of automatic actuation

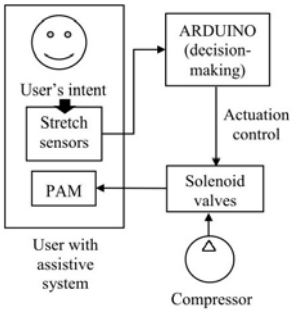

Fig. 2. Block diagram depicting automatic actuation mechanism of PAMs

The artificial muscles inflate and deflate based on the driving air pressure, generated by a commercially available compressor (MEIJI SIRE WHITE 1.25kW). The activation of each set of artificial muscles is controlled by solenoid valves, which are in turn, driven by the resistance information obtained from stretch sensors. Therefore, the real-time data from stretch sensors is used to activate the intended set of PAM(s) and pull the user's hand in the desired direction which completes the motion. This reduces the user's level of muscle activation and eventually, fatigue. The control flow chart of this procedure is given in Fig. 2.

IV. Discussion and conclusions

Force your hand could successfully reduce muscle activation in overcoming passive stiffness, while performing flexion, and extension, in a majority of healthy subjects. However, while performing pronation, the participants showed some degree of resistance when there was a sudden activation of PAM, thus, twisting the wrist. Most of the subjects showed sudden and impulsive peaks in their EMG at the onset of activating the pronation assisting PAM. This indicated some difficulty in overcoming the passive stiffness of wrist while performing pronation in an assisted environment. A change in the current alignment of the PAM to support pronation might be needed to overcome this problem. Apparently, the wrist stiffness varies from person to person for each motion, which could be another reason for the response of assistance in pronation. Fig. 3 illustrates the testing event of force your hand glove on boccia players, who visibly gave positive feedback while using the glove during practice sessions.

Fig. 3. A boccia (paralympic sport) player trying out force your hand glove for wrist empowerment

In our current knowledge, not many efforts have been made in developing soft, wearable, active assist, especially for the wrist of the human body. So, this work is a stepping stone to lay a foundation for developing wearable assist suits for the human's upper limb, having the ability to actively support wrist motions. The evaluation results showed a noticeable reduction in the muscle activity of the majority of participants performing flexion, and extension. This research is not yet content. Future work includes improvement in pronation assistance, testing of the force your hand glove on subjects with wrist muscle weaknesses, and using the glove as a force feedback rehabilitation system by providing motion experience.

References

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Towards the assessment of the functionality and usefulness of robotic extra-finger.

Irfan Hussain[1], Leonardo Meli[1], Gionata Salvietti[1,2] and Domenico Prattichizzo[1,2]

***Abstract*— The use of assessment, clinical or technology-based, is crucial to evaluate the progress along therapy in stroke patients, and to allow monitor usage of assistive and prosthetic devices. Standardized tests that can be used to evaluate both the assistive and rehabilitative facets of robotic extra-finger are still lacking, due to the novelty of the approach. Robotic extra-finger are thought to help patients suffering from sensorimotor hand impairment in their everyday life activities and, eventually, to promote rehabilitation of the paretic arm. Therefore, in the former case, the gold standard for functional assessment is a quantitative comparison of the gain reached in these activities with and without the robotic device. On this line, together with the modification of classical rehabilitation assessments, we propose specific tests based on the execution of activities of daily living (ADL) with use of our robotic extra-finger. In particular, we performed the Frenchay arm test and ADL bi-manual tasks by enrolling six chronic stroke patients. The robotic extra-finger and the paretic hand act like the two parts of a gripper cooperatively holding the grasped object. The subjects were able to perform the tasks that were not possible without the robotic extra-finger.**

Fig. 1. The robotic extra-finger is used to compensate hand grasping abilities for patients suffering from sensorimotor hand impairment).

I. Introduction

Stroke is a leading cause of long-term disabilities, which are often associated with persistent impairment of an upper limb [1]. Findings of available prospective cohort studies indicate that only 5% to 20% of stroke patients with a paretic upper limb manage to fully recover six months after the stroke [2]. However, 33% to 66% show no recovery of upper limb functions after the same period [3]. A key role in functional recovery of stroke patients with a paretic upper limb seems to be played by the improvements of the paretic hand [4]. The recovery of hand functionality also contributes to an increased independence in the Activity of Daily Living (ADL). The use of robotic devices in rehabilitation can provide high-intensity, repetitive, task-specific, and interactive treatment of the impaired upper limb and can serve as a reliable means of monitoring patient's progress [5]. Focusing on the hand therapy, Lum et al. reviewed in [6] several works on robot-assisted approaches to motor neurorehabilitation highlighting the prototypes used in clinical tests. In [7], the authors presented a comprehensive review of hand exoskeleton technologies for rehabilitation and assistive engineering, from basic hand biomechanics to actuator technologies. However, the majority of these devices are poorly wearable, and designed to increase the functional recovery in the first months after stroke, when, in some cases, biological restoring and plastic reorganization of the central nervous system take place. To the best of our knowledge, few devices have been designed to actively compensate hand grasping function when patients are in a chronic state. Indeed, when in the paretic upper limb the motor deficit is stabilized, the rehabilitation mainly consists in ergotherapy, which primarily aims at teaching compensatory strategies by using the non-paretic upper limb and commercial special aids [8]. This potentially increases the functional disparity between the impaired and the unaffected upper limb [9].

Our idea is to augment the functional abilities of the patient with an additional robotic finger prosthesis, the robotic extra-finger, that is worn on the wrist/forearm of the patient, as in Fig. 1. Such a robotic finger is used together with the paretic hand/arm to constrain the motion of the object to be grasped [10]. Due to the complexity of assessing user improvement attributable only to the robot, standards to evaluate the assistive and rehabilitative facets of novel robotic devices are still missing [11]. Currently, the gold standard for functional assessment is a quantitative comparison of the gain reached in certain activities with and without the robotic device. We will discuss how assessment procedures could adapt classical rehabilitation tests to examine assistive devices in the early stage of intense treatment.

The design of the proposed compensatory tools have been driven by robotic and rehabilitation teams, starting from patients requirements in improving upper limb functionality, when the motor deficit is unchangeable [12], [10], [13].

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II. Experiments with chronic stroke patients

Together with the modification of classical rehabilitation assessments, e.g., the Frenchay arm test [14] (as shown in Fig. 2), we propose specific tests based on the execution of activities of daily living [13] (see Fig. 3).

A. The Frenchay Arm Test

Six chronic stroke patients (one female and five males, average age 45) took part to our experimental evaluation. The proposed compensatory tool can be used by subjects showing a residual mobility of the arm. For being included in the experimental phase, patients had to score ≤ 2 when their motor function was tested with the National Institute of Health Stroke Scale (NIHSS) [15], item 5 "paretic arm". Moreover, the patients had to show the following characteristics: normal consciousness (NIHSS, item 1a, 1b, 1c = 0), absence of conjugate eyes deviation (NIHSS, item $2 = 0$), absence of complete hemianopia (NIHSS, item $3 \leq 1$), absence of ataxia (NIHSS, item $7 = 0$), absence of completely sensory loss (NIHSS, item $8 \leq 1$), absence of aphasia (NIHSS, item $9 = 0$), absence of profound extinction and inattention (NIHSS, item $11 \leq 1$).

The goal of the tests was to verify how quickly the patients can learn to use the device and its EMG control interface. We performed a fully ecological qualitative test, the Frenchay Arm Test [14]. The test consisted of five tasks to be executed within three minutes:

1) *Task 1*: Stabilize a ruler, while drawing a line with a pencil held in the other hand. To pass, the ruler must be held firmly.
2) *Task 2*: Grasp a cylinder (12 mm diameter, 5 cm long), set on its side approximately 15 cm from the table edge, lift it about 30 cm and replace without dropping.
3) *Task 3*: Pick up a glass, half full of water positioned about 15 to 30 cm from the edge of the table, drink some water and replace without spilling[1].
4) *Task 4*: Remove and replace a sprung clothes peg from a 10 mm diameter dowel, 15 cm long set in a 10 cm base, 15 to 30 cm from table edge. Not to drop peg or knock dowel over.
5) *Task 5*: Comb hair (or imitate); must comb across top, down the back and down each side of head.

The patient scored 1 for each of the successfully completed task, while he or she scored 0 in case of fail. The subject sat at a table with his hands in his lap, and each task started from this position. He or she was then asked to use his or her affected arm/hand to execute the tasks. Although the Frenchay arm test has not been specifically designed for evaluating compensatory tools, it has shown good reliability in measuring functional changes in stroke patients when comparing with other upper limb assessments [14]. The Frenchay Arm Test contains manipulation tasks of everyday life activity that evaluate the capability of grasping an object without a deep involvement of the patient arm. Other tests like, e.g., ARAT and UEFT [16], are more suitable for the evaluation of arm mobility which is out of the scope of this work.

[1]Note that for safety reasons we did not use water in presence of electronic components.


Task 1: Stabilize a ruler
Task 2: Grasp a cylinder
Task 3: Pick up a glass
Task 4: Remove a sprung
Task 5: Comb hair

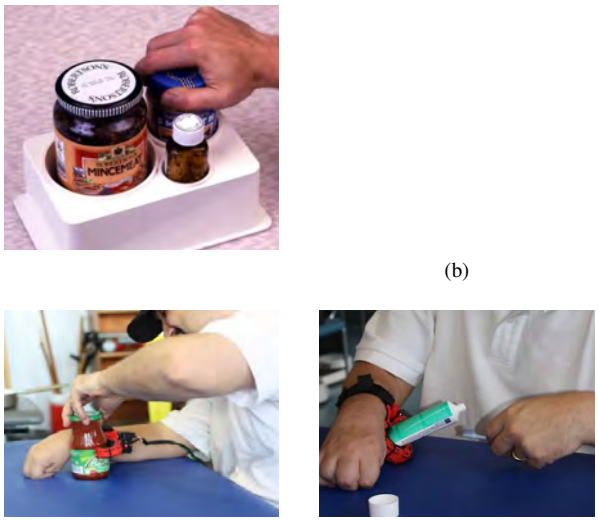
Fig. 2. The five tasks of the Frenchay Arm Test.

The rehabilitation team assisted the subjects during a training phase that lasted for about one hour. During this phase, the optimal position of the device on the arm, according to the patient motor deficit, was evaluated.

After the training phase, the subjects had three minutes to perform the Frenchay Arm Test. Three patients tried the extra-finger for the first time. All the subjects performed the test two times, one without and one with the Soft-SixthFinger. The starting condition was selected randomly. The detailed results are presented in [10], [13]. Screenshots of the tasks are reported in Fig. 2. Note that in the execution of Task 1 the extra robotic finger does not interfere with the paretic limb action. The patients can stabilize the ruler without using any external help, so the device is kept in its rest position.

B. Bimanual tasks

The latter phase of post-stroke rehabilitation is based on the learning of newly acquired motor strategies to compensate the neurological deficit. These strategies may sometimes be neither ergonomic nor ecological, or may even increase pathological motor patterns, usually by worsening tonic flexion at the forearm of the paretic limb [17]. Some compensation techniques take also advantage of dedicated objects that allows the users to execute typical bimanual tasks only with the healthy hand. An example of available aids is reported in Fig. 3. However, the use of such tools is usually limited to the structured houses of the patients, restricting the possibilities of the patients to exploit them



(b) (c) (d)

Fig. 3. Examples of ADL bimanual tasks. On the top, commercial adaptive kitchen aids designed to be used with only one hand in order to perform two specific tasks, i.e., bottle/jar opening and squeezing a toothpaste tube (Pictures courtesy of Elderstore®). On the bottom, a patient performing the same two tasks using only the Sixth Finger.

outside. The extra robotic finger is a portable compensatory tool that can be carried as a bracelet when not used. This allows the patients to bring the devices wherever they want. The regained capability of grasping object with the help of the device, stimulate the patient to use his paretic limb so preserving residual mobility. As a proof of concept, we tested the device in four different bimanual tasks typical of ADL.

In all these bimanual tasks, the paretic limb and the robotic extra finger work together to constrain the motion of the object while the healthy hand manipulate it (e.g., constrain the motion of the tomato jar while the healthy hand is unscrewing its cap). Pictures of the execution of the four tasks are reported in Fig. 3 the patients were able to execute the four tasks without requiring a specific training.

C. Questionnaire

After the Frenchay Arm Test and the bimanual tasks, we investigated the users' subjective satisfaction and possible concerns related to the proposed system. According to [18], questionnaires and interviews are useful methods for studying how users use systems and what features they particularly like or dislike. The patients were asked to fill the Usefulness-Satisfaction-and-Ease-of-use-questionnaire (USE, [19]) that focuses on the experience of the system usage. This questionnaire uses a seven-point Likert rating scale. Mean and standard deviation (SD) of the questionnaire factors are presented in Table I.

TABLE I

QUESTIONNAIRE FACTORS AND RELATIVE MARKS. THE MARK RANGES FROM "1 = STRONGLY DISAGREE" TO "7 = STRONGLY AGREE". MEAN AND STANDARD DEVIATION (MEAN (SD)) ARE REPORTED.

Questionnaire factors	**Mean (SD)**
Usefulness	4.6(0.8)
Ease of use	5.6(0.6)
Ease of learning	6.5(0.8)
Satisfaction	5.8(0.7)

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Design approach of a robotic assistive eating device with a multi-grip and camera for frail elderly's independent life

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Abstract—This project aims at making a mock-up of a multi grip tool for a robotic assistive device and a camera system which enable frail elderly to live more independently and to keep track of their food intake. The robot will be developed through user centred design with analyses of real use cases in Japan and Sweden. Thanks to the collaborating researchers with a strong applied research approach as well as the companies with a strong experience in engineering solutions in both the Japanese and Swedish teams, the functionalities of Bestic, an assistive eating device, could be enhanced for commercial use, and distributed to municipalities and to the general public.

I. Introduction

Undernutrition is a large problem amongst elderly. We believe that it is hard for the user him/herself to notice that they are eating too little. In Sweden many elderly lives alone with the help of home care services and then it is often many different care givers involved, so to keep track of what a person is eating over the day and over time is difficult. These problems occur as well in Japan in the long-term care (LTC) system. On the other hand, Japan and Sweden share the same problem with the demographical development and both countries are robot and technical friendly. This gives a common goal to manage to take care of our elderly 2022.

For most people, chopsticks or knife and fork is sufficient as eating aids. But some need a bit more advanced eating aids due to accident, neurological disease, or congenital disabilities that have caused mobility impairment in arms or hands. Potential users of an eating aid are persons with different neurological disease such as multiple sclerosis, polio, ALS, Parkinson's disease, cerebral palsy, spinal injuries or ataxia.

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Figure 1. The eating aid device Bestic [9].

Up to know, there are several robotic systems already available as outlined in [1-8]. The meal assistance robot My Spoon is characterized by a manipulator arm with 5-DOF and an end-effector that is controlled by a joystick, with a spoon of 1-DOF [1]. The assisting robotic system Handy 1 has a 5-DOF robotic arm that is installed on a non-powered wheeled platform to assist in very specific activities of daily life such as eating, drinking, and make-up application [5]. During the eating application a scanning system of lights has been included into its tray section which helps the user to select food from any part of the dish.

The eating aid, Bestic, is made to get the food from the plate to the mouth, controlled by the user (Figure 1). The development of Bestic started as a master thesis by Ann-Louise Lindborg at Chalmers in 2004 [10]. Meanwhile, Sten Hemmingsson has stabilised a company around the product, called Bestic AB. The design concept, where contact with the users during the whole development process was considered [9], was focused on getting friendly, kitchen-aid looking shape with the following characteristics: small, quiet and portable.

In our project, we are aiming to enable Bestic to create food intake reports in order to decrease undernutrition among frail elderly by collecting data via a vision system connected to the Internet. For this purpose, the authors have proposed the following objectives:

- Adapt the eating assistive robot to Japanese customs and meals.
- Create food intake reports via a vision system connected to the internet.

- Provide new possibilities for independence by adding new functionalities with multi-grip tools.

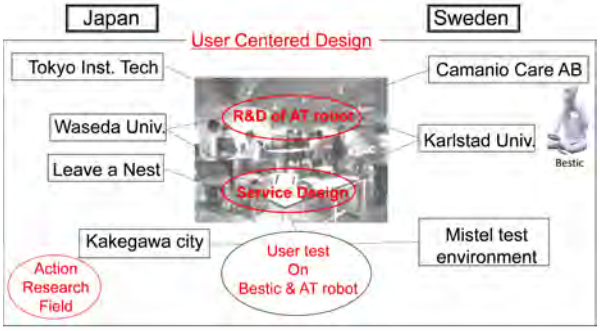

Figure 2. Overall international organization and approaches to the proposed project.

In this project, the authors intend to introduce robotic devices to keep frail elderly's independence of life and also save families or care professionals time and reduce psychological burden for frail elderly care. To be independent and not need to ask for help or describe how the help/work shall be done is highly valued by most persons. We have a vision for aging society which support and widen the possibility of their independent life by introducing novel multimodal and multipurpose robotic assistive device with multi-grip tools and vision system.

For this purpose, the authors have proposed the following possible approach (Figure 2):

- User centred design and challenge oriented approach
- User tests and interviews to gather data about: the whole mealtime situation, how the assistive eating aid works, how caregivers and relatives are affected and what more do they wish to use the assistive aid for (except for eating for the users) testing Bestic in Japan and Sweden.
- Agile prototyping of safe multi-grip tool for the eating assistive device in order to widen the usability in different ways.
- Make a feasibility study of the vision system and test different potential applications with innovative technology.

A vision system that is possible to connect to Bestic opens up much possible functionality. The aim is to make a stand-alone-system that can collect data of how much a person has eaten. It will be connected through wireless communication and to send pictures and data. We can also use the camera system for defining where on the plate the food is.

To further develop the Bestic arm with a multi-grip-tool that can lift different things. By agile prototyping that pseudo users can test we believe that new needs will be possible to define. We propose implementation system for both societies based on the investigation on perception of Bestics and regulation barriers in order to create prospective services using robotic assistive devices.

III. Preliminary Findings

Can we include something outcome from our discussions in Stockholm or some preliminary studies in Japan?

After Japanese and Swedish teams joint meetings in Stockholm, Sweden in March 2017, we have the following preliminary findings.

- User test is very important for the success of this project. We can find the possible problems of our proposed system as well as user's needs.
- We need to adjust food preparation on a plate in each country owing to their big differences in eating culture. The Japanese food styles are very different from that of Sweden.
- The related ethical regulations in Japan and Sweden for users' personal data should also be clarified in the early stage of the project
- As well as the technical development, design of user's experience through newly developed Bestics is quite important. This will be step-by-step clarified through design-prototyping-user test-market research.

IV. Conclusion

In this paper, the approach for the proposed development for introducing novel multimodal and multipurpose robotic assistive device with multi-grip tools and vision system has been introduced, and we also have summarized the preliminary findings on user test, culture differences and ethical regulations

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Proposal of Exoskeleton with Variable Viscoelastic Joint System named "Airsist I" *

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I. Introduction

Recently, several human assisting devices have been developed to reduce labor-intensive workloads and support individuals with disabilities [1-2]. The effectiveness of such devices has been confirmed by their implementation in diverse applications, such as in reducing surface EMG.

The exoskeletons need to be driven softly to provide the wearer with safe and cooperative assistance. However, existing human-assisting devices driven by motors and reduction gears can achieve only superficial softness through the control of torque and joint angles. This method requires many sensors, thus making the devices expensive. In addition, existing actuation methods cause degradation of energy efficiency, because motors, sensors, and gearboxes are needed at the movable members, resulting in high energy consumption for the movement of the device itself. Moreover, the method restricts the natural motion of the wearer since the actuator has low backdrivability during power source cutoff.

On the other hand, human body joints are moved and controlled by variable viscoelasticity generated by antagonized muscles [3]. Therefore, as discussed above, the actuation method of human body joints differs from other actuation methods such as torque and angle control. Using an assisting device having the same actuation method as that of a human body joint in parallel with the human body joints is expected to provide high compatibility of the exoskeleton with the wearer. However, assisting devices of this type have not been developed to date.

Thus, we have been studied a variable viscoelastic joint system composed of pneumatic artificial muscles and magneto rheological brakes [4-5]. It was validated that the proposed system can be driven softly without the need for sensors and provides high backdrivability for the wearer. However, experiments in previous study are conducted using 1 degree of freedom prototype. In this paper, a lower body exoskeleton named "Airsist" is developed in order to evaluate the proposed method in practical conditions.

II. Concept of the Proposed Viscoelastic Joint

The concept of the proposed actuation system is illustrated in Fig. 1. It consists of antagonized artificial muscles and MR brakes. The artificial muscles that are attached to the input shaft through a pulley help in achieving variable elasticity. The joint angle, torque, and elasticity can be controlled independently by controlling the air pressure of the two artificial muscles without using feedback control. Although an air cylinder may also be used to effectively achieve variable elasticity, it needs a feedback control for the angle because the contractive force of the air cylinder is constant at constant pressure. Pneumatic actuators have advantages that they are lightweight and structurally flexible; however, they also suffer from slow response, which causes response delay in the system. A schematic diagram of the pneumatic artificial muscle is shown in Fig. 2. The artificial muscle is composed of carbon fibers and latex rubber and has a construction force greater than that of the Mckibben-type artificial muscle, which is generally used as artificial muscle [6].

Variable viscosity can be achieved by controlling the friction torque using the Magneto Rheological brake (MR brake) according to the rotation speed of the input shaft. MR brake is a device which utilize quick response of MR fluid.

The proposed system provides high backdrivability owing to the elasticity of the actuator and the slackness of the wire.

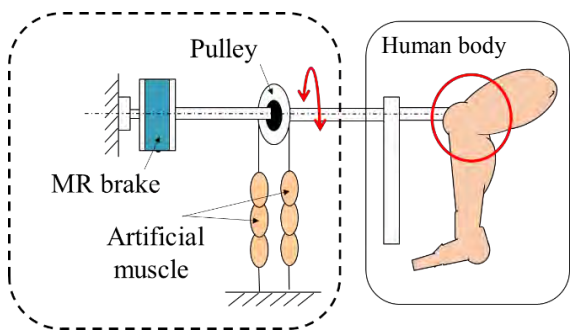

Figure 1. Variable viscoelastic joint system.

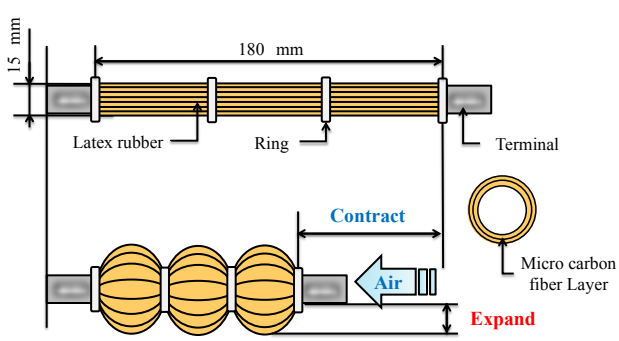

Figure 2. A schematic of the pneumatic artificial muscle.

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III. Lower Body Assistive Exoskeleton

The schematic of the design of Airsist is shown in Fig. 3, and Table I shows its specifications. Airsist is an exoskeleton that have variable viscoelastic joint for assisting hip and knee joints. It also has a passive ankle joint, and the ankle part consists of a plate that contacts ground, which transmits some load of Airsist's weight to the ground. The wearer can wear general shoes.

The feature of Airsist is lightweight design owing to adoption of high-power-density pneumatic artificial muscles and MR brakes. Knee and hip joints can demonstrate a maximum of 88 Nm assistive torque; this renders Airsist capable of assisting up to about 50 % of the required torque for a 70 kg adult male in a squatting motion.

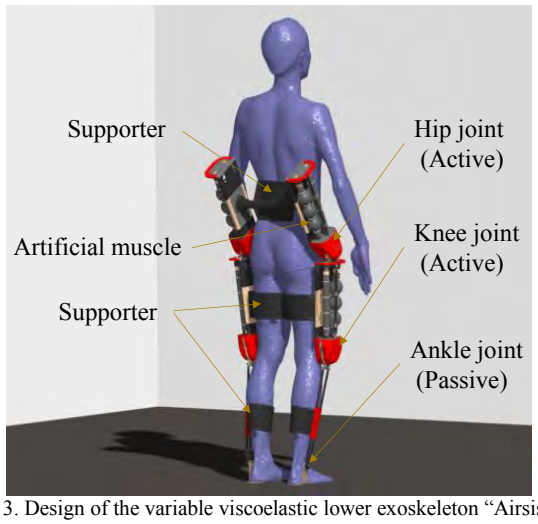

Figure 3. Design of the variable viscoelastic lower exoskeleton "Airsist".

TABLE I. SPECIFICATION OF THE VARIABLE VISCOELASTIC LOWER EXOSKELETON "AIRSIST"

Whole Weight	7.9 kg
Weight of leg	1.9 kg
Range of motion (hip)	Extension 5°, flexion 120°
Range of motion (knee)	Extension 20°, flexion 80°
Assistive force	88 Nm(at maximum, artificial muscle is in isometric condition)
Pulley radius	20 mm

IV. Test Using A Prototype

An image of the assistive experiment is shown in Fig. 4. The proposed joint system is found to be able to assist according to human movement. During the assistance, unexpected vibration is not observed. The result shows that the variable viscoelastic joint system and its controller can follow human movement.

Fig. 5 shows images of backdrivability experiments. As shown in Fig. 5, the prototype provided high backdrivability while the wearer climbed stairs. Also, it is confirmed that the wearer could run at the speed 10 m/s in a running experiment, owing to the low inertia provided by Airsist.

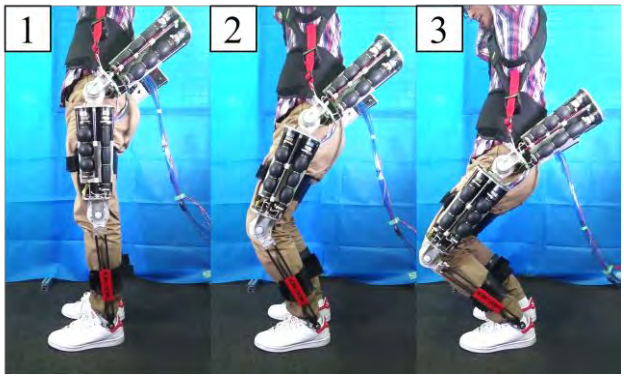

Figure 4. Image of assistive experiment.

Figure 5. Image of stair climbing experiment.

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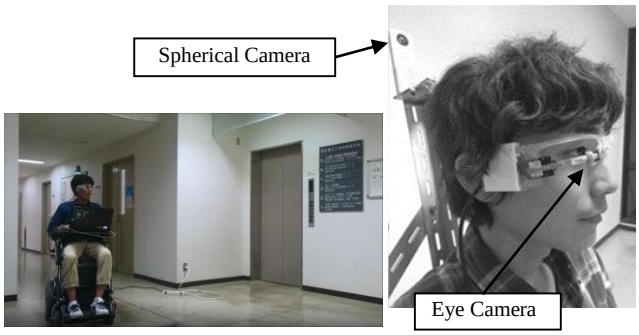
Gaze-Based Interface for Wheelchair Robot - Acquiring Gaze Trajectory by Full-View Camera*

Shigang Li, *Member, IEEE*, Tatsuya Fujiura, and Isao Nakanishi, *Member, IEEE*

I. Introduction

How to sense human's intention is a basic and important problem for the construction of a more seamless interface between humans and machines [1]. In this paper, we focus on the cues of human's gaze. For a user sitting on a wheelchair, the objects a user gazes at maybe be located at the whole environment surrounding him/her. To analyze his/her gaze action, we need to know the his/her gaze trajectory in the environment map.

Figure 1(a) shows a concrete case considered in this research. A user sitting on a wheelchair moves at an indoor environment. If we know his gaze trajectory at the room, we can support him more effectively and more efficiently. The problem is how we can obtain the user's gaze trajectory in such an environment.



(a) (b)

Figure 1. (a) A user sitting on a motor chair moves at an indoor environment. (b) The proposed system consisting of a spherical camera and an eye camera.

A natural and simple method to answer the above question is to detect the user' gaze point and plot it in an environment map. Thus, the problem becomes how to detect a user' gaze point, how to acquire an environment map, and how to plot the detected gaze point at the acquired environment map.

Figure 1(b) shows the system proposed in this paper. A wearable eye camera is used to acquire a corneal reflection image, and the user' gaze point is detected in the corneal reflection image. A full-view spherical scene camera, which is mounted on a wheelchair robot, is used to acquire the surrounding environment map. By mapping the gaze point detected in the corneal reflection image onto the environment map, the gaze trajectory is recorded in the environment map. By analyzing the recoded gaze trajectory, the wheelchair robot can understand the user's intention. In this paper, we focus on how to acquire a user's gaze trajectory for an omnidirectional-vision wheelchair robot.

II. Acquiring Gaze Trajectory for Wheelchair Robot

Figure 2 shows the block chart of the proposed algorithm. The algorithm is divided into 3 parts: preprocessing of full-view image, processing of eye image and mapping of gaze points, as indicated by the enclosed red dot lines. Next, we give the details of the proposed algorithm.

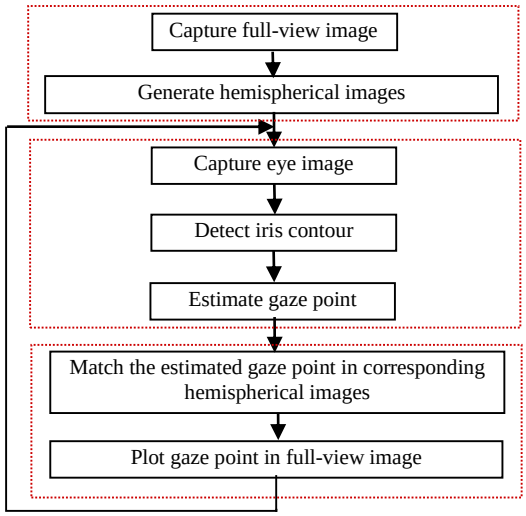


Figure 2. Block chart of the algorithm proposed in this paper.

A. Processing of Full-View Image

Figure 3 (a) and (b) show a full-view image captured by the full-view scene camera mounted on the motor chair, and a corneal reflection image captured by the head-mounted eye camera, respectively; as shown in Figure 1(b).

B. Estimation of Gaze Point

Here we explain how to estimate gaze points from the corneal reflection image captured by an eye camera [1]. First, we roughly locate the center of the iris of the eye, which is based on the gradient information of edge points. Using the center information, a limited region enclosing the iris of the

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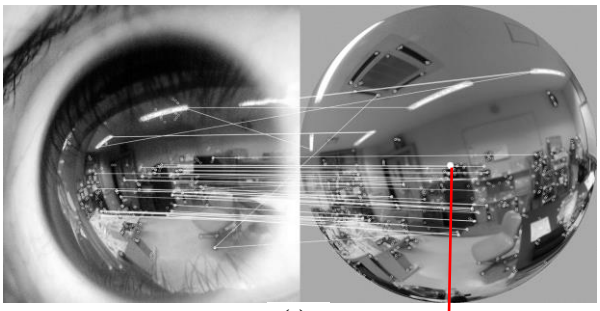
(a) (b)

Figure 3. (a) A full-view image captured by a spherical camera. (b) A corneal reflection image captured by an eye camera.

eye is cut out. Then, a median filter is used to delete the noise and blur the image so as to have a better iris contour fitting. Next, the Canny operator is used to extract the candidates of edge points on the iris contour. Furthermore, a RANSAC algorithm is used to fit the iris contour. Finally, the point of gaze is estimated from the corneal reflection image.

C. Mapping of Gaze Point

At first, the corneal reflection image is matched with the most similar hemispherical image of the 26 images generated in Subsection 3.1. First, the brightness of the two images is normalized by a histogram-equalized technique. Next, AKAZE [3] features are extracted from the two images, respectively; and the robust matching is carried out by a double check technique, as shown in Figure 4(a). Then, the rotation matrix between the two cameras is computed using the matched features. Finally, the computed rotation matrix is used to map the gaze point from the corneal reflection image to the full-view image, as shown in Figure 4(b).



(a)

(b)

Figure 4. (a) Matching AKAZE features between the corneal reflection image and hemispherical image generated from full-view image to compute their rotation matrix. (b) Mapping the gaze point to the full-view image by the computed rotation matrix.

III. Experimental Results

Here we show a preliminary experimental result on controlling a wheelchair robot by gaze action. As an example, we show how to control a wheelchair robot to make a turn at a junction in indoor environment. The strategy is as follows.

- Two coordinate systems are set at the proposed system: one corresponds to the wheelchair robot, $O-XYZ$, while the other corresponds to the spherical camera mounted on the wheelchair, $W-UVS$, as shown in Figure 5. The relative pose between these two coordinate systems is calibrated beforehand.
- Next, the gaze point of the user is estimated from the eye camera, and is mapped on the full-view image by the method described in Section 2.
- Then, according the position of the estimated gaze points in the full-view image, one of the three movement patterns (move forward, turn left, turn right) is selected. The demo video of this preliminary test is available at [4].

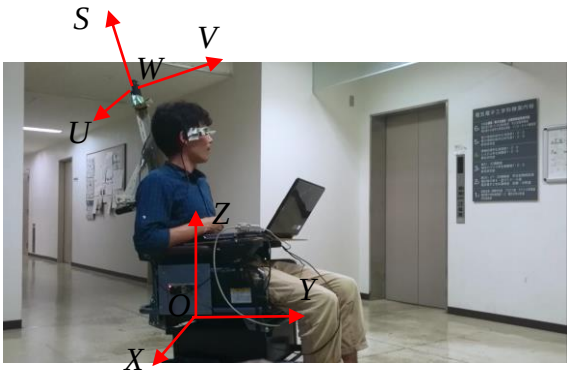


Figure 5. Two coordinate systems used in our spherical vision wheelchair robot.

IV. Conclusion

In this paper, we focus on the recording of the gaze points by a spherical camera mounted on a wheelchair and give the preliminary experimental result on controlling the motion of the wheelchair by gaze action. How to analyze the intention of users according to the trajectory of gaze points and control a wheelchair robot smoothly based on a gaze interface are our future work to do.

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A Networked Control System Using Wi-Fi Communication and Its Application to Assist Cart Control*

Susumu Hara, *Member, IEEE*

***Abstract*— The recent increase in the amount of aged citizens in the society has prompted the development of power-assist systems. The research group of the author has previously investigated the control system of robust assist carts such as those used in factories. However, the heaviness of such carts—due to the installed high-specification computers and lithium-ion batteries—makes them unsuitable for welfare purposes such as walking assistance of the aged. In the present study, the authors developed a novel network control system for use in robust power-assist carts for welfare applications. The developed system involves the use of a Wi-Fi network control configuration to connect an external high-specification computer to a simpler computer installed on the cart. This enables significant weight and cost reductions in the cart while retaining the computational control ability. The stability of the proposed cart with and without Wi-Fi control was experimentally confirmed.**

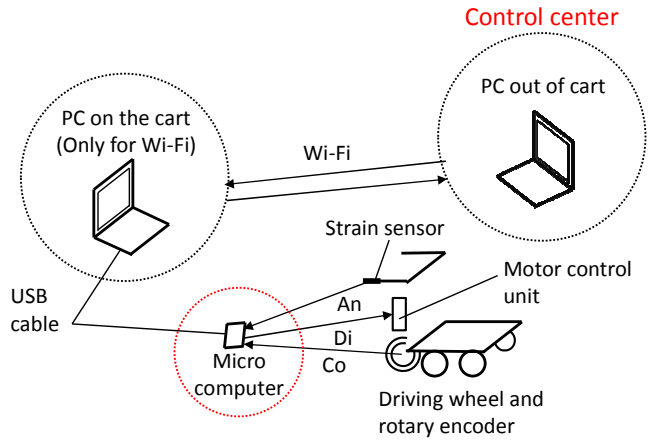

Fig. 1. Wi-Fi remote control system.

I. Introduction

The recent increase in the amount of aged citizens in the society has prompted the development of power-assist systems. Generally, a power-assist system increases the operating power of the user by means of motors, based on the forces applied by the user as determined by sensors. The research group of the author has previously discussed robust assist systems for suppressing the vibration and preventing the tumble of conveyed objects [1], [2]. These objectives were achieved in [1] by equipping the controlled object, namely, an assist cart, with two types of actuators. The actual control system consisted of an impedance controller, a disturbance observer, a reaction force controller, and a frequency-shaped disturbance accommodating optimal controller. The effectiveness of each element of the control system was verified by experiment and simulation.

The assist cart in [1] was equipped with two lithium-ion batteries, which were used to power installed high-specification control devices. Considering that such carts are mainly used to support production activities, their operation conditions are limited. However, they are heavy and would require suitable tuning to deliver ideal assistance under more diverse operation conditions. Assist carts used for household and welfare purposes require such adaptability and there is the need to minimize their weight.

To this end, this paper proposes the use of a networked control system. Recently developed wireless technologies have facilitated the use of network systems to solve various control problems [3]. For example, Kohinata et al. proposed a wireless network control system for a rotary inverted pendulum [4], and used simulation to determine the effects of the channel errors on the control system. Another relevant development is a teletherapy surgery-support system [5]. Indeed, network control affords diverse innovations.

The operation of the presently proposed system involves five steps, as shown in Fig. 1. Firstly, a microcomputer on the assist cart receives data from the sensors, such as about the displacement of the cart and the operating force, and the data is transmitted to the cart computer. Secondly, the data is transmitted to an external computer (the control center) by Wi-Fi. Thirdly, the control center uses the data to generate the control input based on a control theory. Fourthly, the control input is transmitted to the cart computer, and, finally, to the microcomputer, which executes the assist control. This control configuration eliminates the need to install a high-

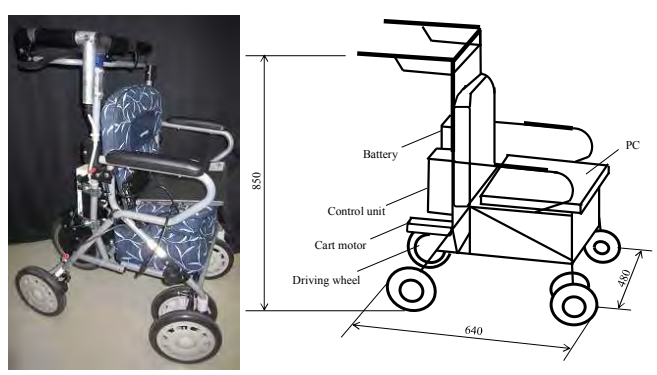

Fig. 2. Controlled object.

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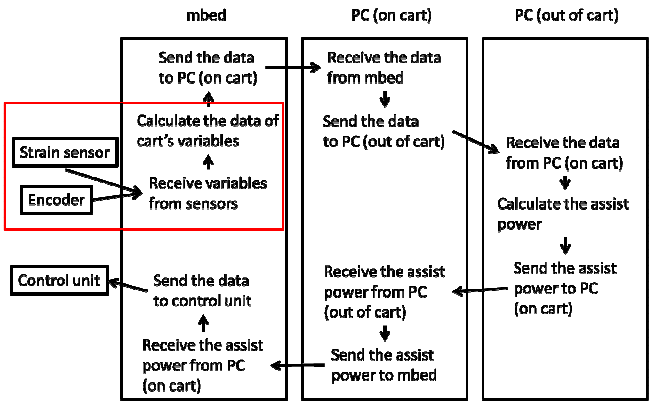


Fig. 3. Flowchart of the proposed system.

specification computer and high-capacity batteries on the assist cart. It thus significantly reduces the weight of the cart while retaining the computation ability of the control system. The employed cart computer is used for network communication and power supply to the microcomputer. Thus, it is possible to replace the battery and the network communication tool with the cart computer. In that case, the cart computer is not required.

The application of the proposed network control configuration to welfare assist carts offers some advantages. Firstly, it enables monitoring of the state of the cart through the control center. This makes it possible to collect information about, for example, hazardous areas, well-used roads, and traffic volume. The authors of [6] highlighted some related issues regarding the use of a rollator, and noted that the acquisition of such information generally requires hearing surveys. However, the presently proposed system simplifies the information acquisition process. In addition, the information can be used to improve the convenience and safety of all users of similar systems. Another advantage of the proposed system is that it enables access to the control center, thus facilitating modification of the contents of the control system. This means that, if a problem arises in the control system, another control system and new parameters can be transmitted from the control center to solve the problem. The configuration thus affords more efficient maintenance and tuning of the system parameters.

The present study focused on a walking assist cart for the aged, known as a rollator. Figure 2 shows a photo and schematic drawing of the particular Japanese rollator considered in this study. Unlike a wheel chair, a rollator helps the user to walk. From the viewpoint of the user's health, it is a useful piece of equipment. When equipped with a power-assist systems, a rollator enables the user to walk on terrains that are typically difficult for aged people, such as slopes and steps. In [7], the authors presented the relationship among the operating force of the cart, the height of the step, and the diameter of the cart wheels, and showed that it was difficult to use a rollator without power assist to climb a step. The application of power assist to the cart thus enables the user to significantly expand their area of activity.

The concept of the proposed system was first presented by the author in an IEEE conference paper [8]. The present paper is an expanded version of this previous work. The control system configuration was improved in the present work through the use of an mbed™ microcomputer, which smoothened all the motion trajectories.

II. Controlled Object

The controlled object of the present study is shown in Fig. 2. The assist cart is made of a silver cart on the market. A strain gauge is attached to the right handle of the cart to measure the operating force of the user. The driving wheel is also equipped with a rotary encoder that measures the displacement of the cart. A microcomputer (ARM® mbed™, CPU: ARM Cortex-M3 LPC1768 96 MHz, FLASH: 512 KB, RAM: 64 KB) is connected to the cart computer and used to transmit and receive data. As explained in the introduction section, the data acquired by the strain gauge-type sensor and rotary encoder are transmitted to the control center by Wi-Fi, where they are processed by a control algorithm. The generated control input is then transmitted back to the mbed microcomputer for appropriate powering of the cart.

III. Control Method

A. Network control system

The flowchart of the proposed network control system for the experiments in the following section is shown in Fig. 3. The operation of the system consists of seven steps. Firstly, the microcomputer receives data, for example, regarding the displacement of the cart and the operation force applied by the user, as measured by the employed sensors. Secondly, the data are transmitted to the cart computer. Thirdly, the data are subsequently transmitted to the control center (PC outside the cart) by Wi-Fi. Fourthly, the control center uses the data to generate the control input based on a control theory. Fifthly, the control input is transmitted to the cart computer. Sixthly, the control input is subsequently transmitted to the cart microcomputer. Finally, the microcomputer executes the assist control. The sampling period of the network control system is approximately 100 ms. The measurement process in the red square in Fig. 3 is executed at intervals of 10 ms.

B. Assist control algorithm

To adequately achieve the assist, an impedance control algorithm [9] is applied to the control. The impedance controller is implemented in the same manner as in [1], [2]. The control system moves the assist cart such that its mechanical impedance parameters such as the mass of the cart and the coefficient of viscous damping are appropriately set. The impedance controller utilizes the following equations:

$$m_w \ddot{x}_w(t) = f_{imp}(t) + f_h(t) \tag{1}$$

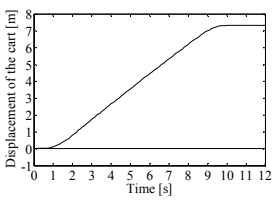
$$m_{id} \ddot{x}_w(t) + c_{id} \dot{x}_w(t) = f_h(t) \tag{2}$$

$$f_{imp}(t) = -\frac{m_w}{m_{id}} \times c_{id} \times \dot{x}_w(t) + \left(\frac{m_w}{m_{id}} - 1 \right) \times f_h(t) \tag{3}$$

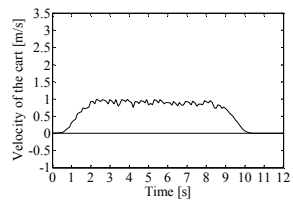
Equation (1) describes the linear motion of the assist cart, while Eq. (2) describes the dynamics of the impedance characteristics based on the impedance parameters.

TABLE I. PARAMETERS OF THE CONTROLLED OBJECT

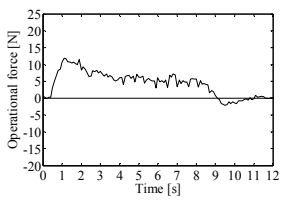
Symbol	Parameter	Value
m_w	Mass of the cart	22.2 kg
m_{id}	Ideal mass of the cart	7.0 kg
c_{id}	Ideal viscous damping coefficient of the cart	5.0 N・s/m



(a) Displacement of the cart



(b) Velocity of the cart



(c) Operating force

Fig. 4. Experimental results without impedance control.

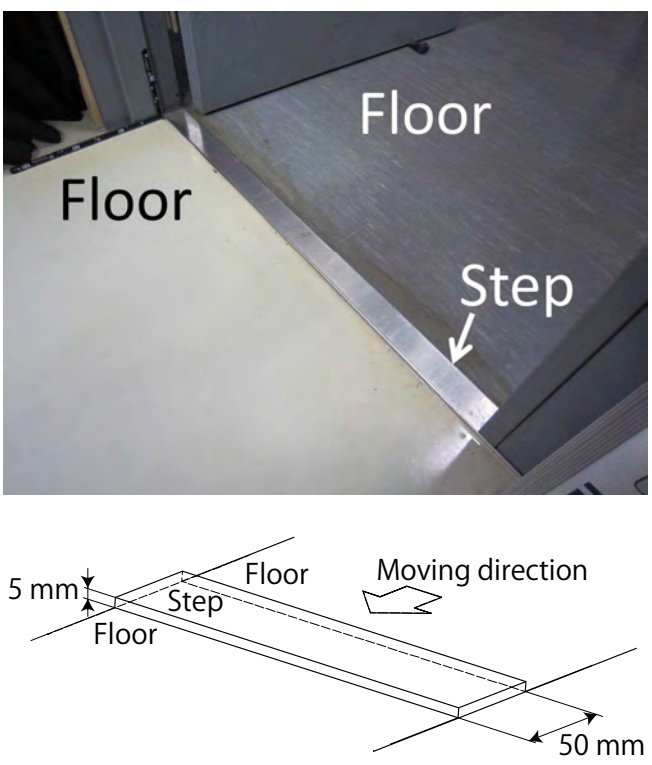


Fig. 5. Picture and specifications of the step.

Equation (3) corresponds to the impedance control input used to achieve the dynamic characteristics in Eq. (2). $x_w(t)$ and $f_h(t)$ respectively represent the displacement of the cart and the operating force applied by the user, while $f_{imp}(t)$ represents the thrust force of the cart determined by the impedance control algorithm. The impedance parameters are specified in Table 1. In this study, m_{id} was set to about one-third of the original m_w. The use of an appropriate c_{id} enables easy positioning and ensures control stability when the cart moves within a small range.

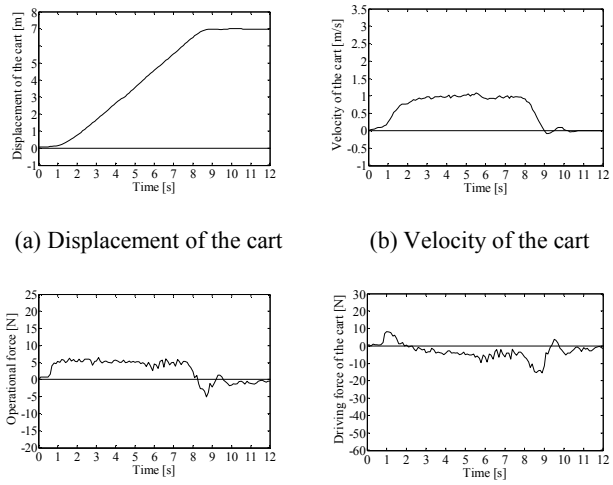
To ascertain the validity of the impedance control, we conducted experiments using and without using the impedance control. In the experiments, the cart was operated on a flat road over a distance of about 7 m for about 9 s. Figure 4 shows the experimental results for the case without impedance control, while Fig. 6 shows the results for the case with impedance control. As can be observed from the figures, the maximum operating forces for the two cases are 11.9 and 6.5 N, respectively. This shows that the force required to operate the cart without impedance control for a given velocity is approximately 1.8 times that required to operate the cart with impedance control. This confirmed the effectiveness of impedance control.

Because the present study was a fundamental one with the primary objective of demonstrating the application of network control to a power-assist cart, only impedance control was considered. The utilization of a high-specification control center would enable the network control system to implement advanced control computations that the cart computer is incapable of. The author plans a future report on the implementation of such a system.

IV. EXPERIMENTS

A. Details of experiments

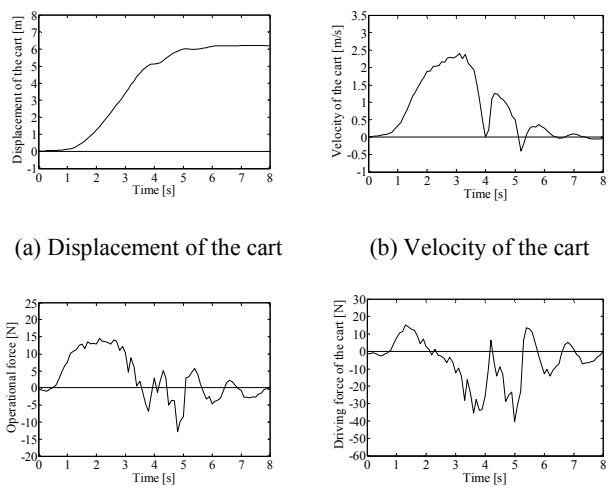
To examine the performance and stability of the proposed Wi-Fi network control, for experimental cases were performed. In Case 1, the assist cart was operated on a flat road. Case 2 also involved a flat road, but the initial acceleration and final deceleration were higher than in Case 1. In Case 3, the assist cart was operated on a step. Figure 5 shows a photo and schematic of the step. The height and width of the step were 5 and 50 mm, respectively. The assist cart climbed the step in approximately 4 s. Figures 8 and 12 show the variations of the operation parameters with time. The increase in the operating force is due to the cart user having to push harder to climb the step. Case 4 of the experiment involved the assist cart colliding with the wall three times. The stability of the network control system during sudden acceleration and the application of external forces was examined in these experiments. For comparison, all the experimental cases were performed with and without using the Wi-Fi network control system. When the Wi-Fi system was not used, impedance control was implemented through the cart microcomputer. To facilitate the comparison, the sampling period of the experiments without process was implemented at intervals of 10 ms.



(a) Displacement of the cart (b) Velocity of the cart

(c) Operating force (d) Driving force of the cart

Fig. 6. Experimental results for the implementation of the proposed Wi-Fi network control (Case 1: flat road).



(a) Displacement of the cart (b) Velocity of the cart

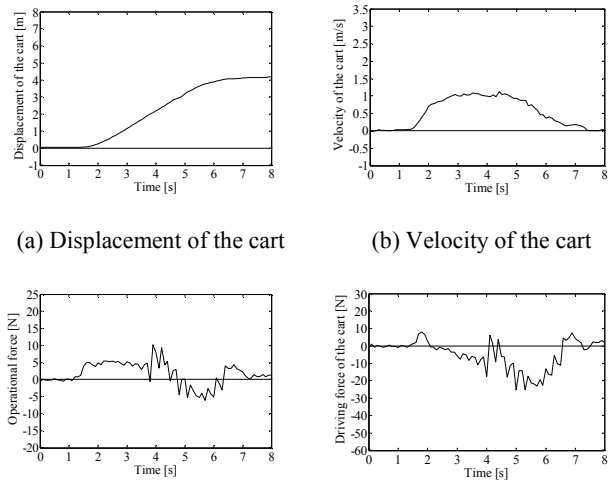
(c) Operating force (d) Driving force of the cart

Fig. 7. Experimental results for the implementation of the proposed Wi-Fi network control (Case 2: flat road with higher acceleration and deceleration).

B. Results of experiments using and without using Wi-Fi control

The results of the experiments in which the proposed Wi-Fi network control were used for Cases 1, 2, 3, and 4 are shown in Figs. 6, 7, 8, and 9, respectively. The corresponding results of the experiments performed without using the Wi-Fi control are shown in Figs. 10, 11, 12, and 13, respectively.

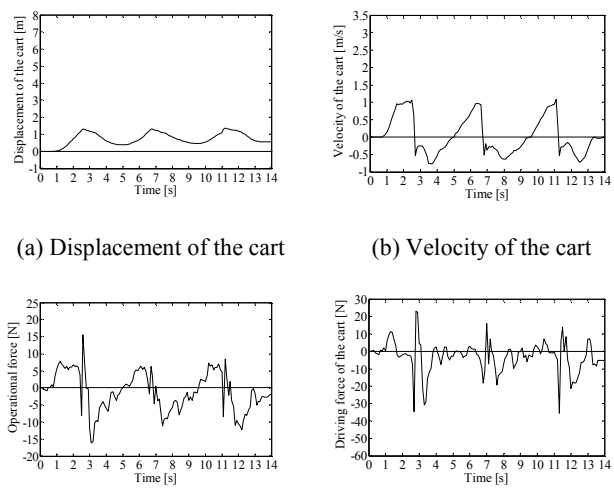
As can be observed from the figures, the corresponding results obtained with and without Wi-Fi control are similar. This verifies the effectiveness of the proposed Wi-Fi system, including in real environments.



(a) Displacement of the cart (b) Velocity of the cart

(c) Operating force (d) Driving force of the cart

Fig. 8. Experimental results for the implementation of the proposed Wi-Fi network control (Case 3: step climbing).



(a) Displacement of the cart (b) Velocity of the cart

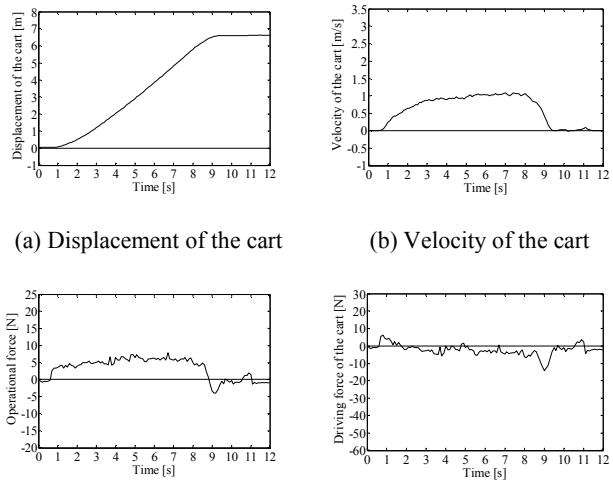
(c) Operating force (d) Driving force of the cart

Fig. 9. Experimental results for the implementation of the proposed Wi-Fi network control (Case 4: collision with wall).

V. CONCLUSIONS

In this paper, we discussed the application of a novel Wi-Fi network control system to a Japanese rollator. The controlled object and control method were described, as well as experiments that were used to verify the effectiveness of the control system. An adequate foundation was confirmed to have been laid for the practical application of the proposed system.

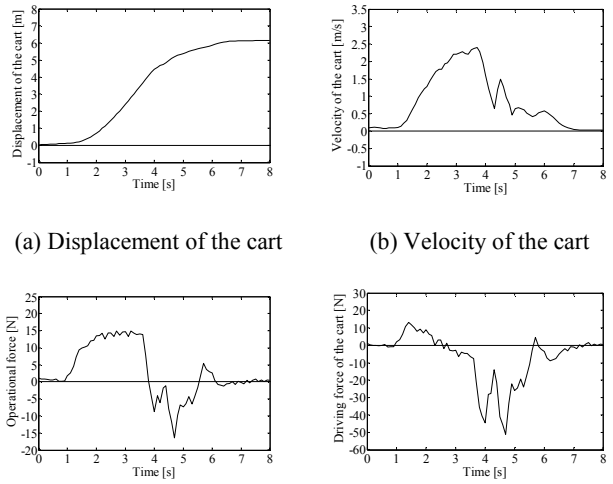
The direction of our further study is four-fold. Firstly, we plan to introduce a multi-rate control system [10] that applies different sampling periods to the cart computer and control center, respectively. Further, as indicated by the results of the present experiments performed using and without using the



(a) Displacement of the cart (b) Velocity of the cart

(c) Operating force (d) Driving force of the cart

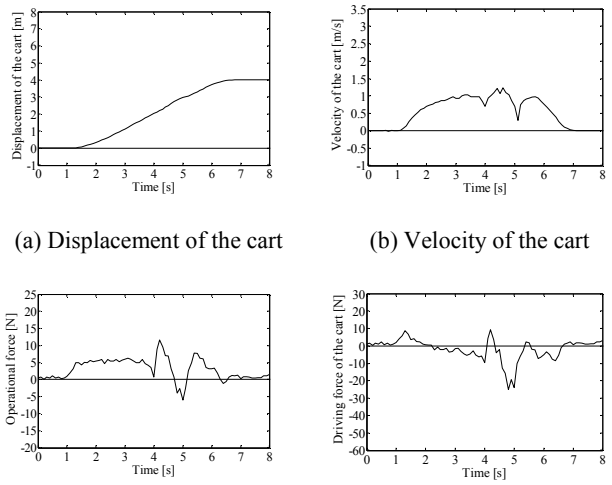
Fig. 10. Results of the experiment without using Wi-Fi control (Case 1: flat road).



(a) Displacement of the cart (b) Velocity of the cart

(c) Operating force (d) Driving force of the cart

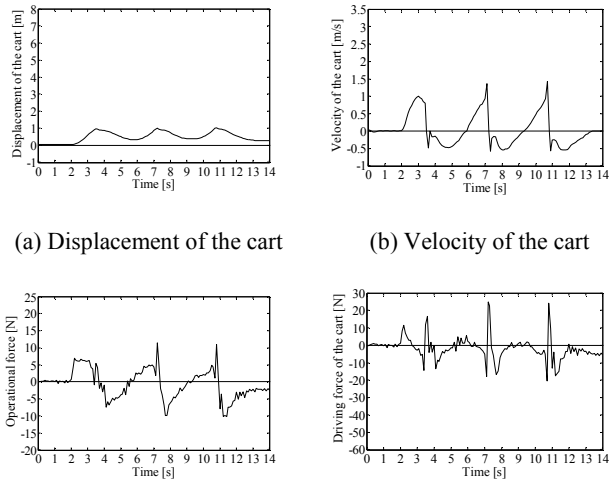
Fig. 11. Results of the experiment without using Wi-Fi control (Case 2: flat road with higher acceleration and deceleration).



(a) Displacement of the cart (b) Velocity of the cart

(c) Operating force (d) Driving force of the cart

Fig. 12. Results of the experiment without using Wi-Fi control (Case 3: step climbing).



(a) Displacement of the cart (b) Velocity of the cart

(c) Operating force (d) Driving force of the cart

Fig. 13. Results of the experiment without using Wi-Fi control (Case 4: collision with wall).

Wi-Fi control, the proposed system is not ideal for an equipment requiring fast control. The envisioned multi-rate control system is expected to afford faster, better, and more intricate control. As a first step toward achieving this control system, the network control and data sampling periods were separated in the present study.

Secondly, we plan to integrate other control elements with the impedance controller, as was done in [1]. Thirdly, we will investigate the safety of the feedback network control system, as was done in [5]. For example, we will consider the effects of the unpredictable uncertainties of a real system [11]. It is noteworthy that the safety of the entire feedback network control system was not considered in the majority of previous studies. Fourthly, to make the best of the proposed remote control system, we will consider a more intricate control law for the control center, such as one that enables trajectory generation for optimal control.

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Development of a dynamically stable wheelchair of single spherical drive

Tasuku Hoshino[1], Miki Yazawa[1], Hiroshi Miyazawa[2] and Hiroshi Kobayashi[2]

***Abstract*— A dynamically stable wheelchair driven by a single sphere is proposed, designed, and implemented in this study. The sphere drive structure realizes immediate mobility and dynamical stability to all directions and the ability to turn on a point. The leaning-based operating method of the wheelchair provides a means of physical rehabilitation for the user. The dynamical stability also helps the attendant to roll the wheelchair with a bit of effort even on a rising slope, which prevents him/her from damaging his/her back. The system descriptions, the modeling and the controller design procedures are shown and experimental results are also included.**

I. INTRODUCTION

In the coming aging society, wheelchairs have growing importance in nursing care both at home and at the hospital. Conventional wheelchairs, whose basic structure had been proposed more than 200 years ago, have a large driving wheel on each side of an armchair and two small assisting wheels in front, which make the wheelchair statically stable. However, the stability margin becomes smaller on a sloping surface or when the acceleration is large. The problem arises regardless whether the wheelchair is driven manually or electrically.

Dynamically stable wheelchairs such as Genny [3] and iBot [4] have been developed in recent years. Although they are unstable in nature since there is no assisting wheels and they are standing with two parallel driving wheels, appropriate feedback control achieves the dynamical stability and realizes smooth acceleration. It maintains stability on uphill or downhill slope by keeping the vehicle body upright. However, despite the significant merit of dynamical stability, this property is limited in front-back direction because of their two-wheel structure.

This paper proposes a dynamically stable wheelchair driven by a single sphere, which realizes immediate mobility and the dynamical stability to all directions (Fig. 1). The width can be reduced since there is no need of driving wheel on both sides, and moreover, the wheelchair is able to turn on a point; these are important advantages when going through a narrow lane especially in a house. The wheelchair can be driven by leaning user's upper body; it provides a means of physical rehabilitation for the user. The dynamical stability also helps the attendant to roll the wheelchair with a bit of effort even on a rising slope, which prevents him/her from damaging his/her back. The authors named the wheelchair "Spheel."

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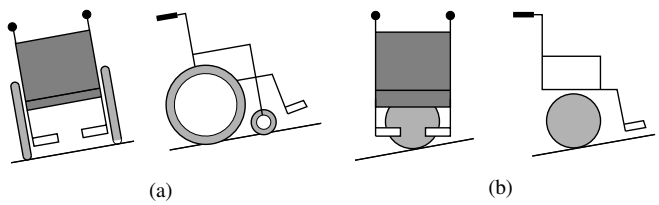


Fig. 1. Difference of mechanical stability: a) static stability of conventional wheelchair and b) dynamical stability of wheelchair of single spherical drive; the former degrades on a sloping surface while the latter is not affected.

Fig. 2. Prototype of a dynamically stable wheelchair of single spherical drive (Spheel)

II. DESIGN AND DESCRIPTION OF WHEELCHAIR

Figure 2 is the proposed wheelchair, consisting of an armchair and a driving sphere. Four omni wheels driven by AC servomotor through gear train are mounted under the seat, in order to rotate the sphere about three orthogonal axes (Fig. 3). The sphere is set in contact with the omni wheels firmly using other four free omni wheels. There are also equipped four supporting legs at each corner of the chair. These legs vertically slide down to fix the chair on the floor when the user get on and off; they slide up when the wheelchair is in operation.

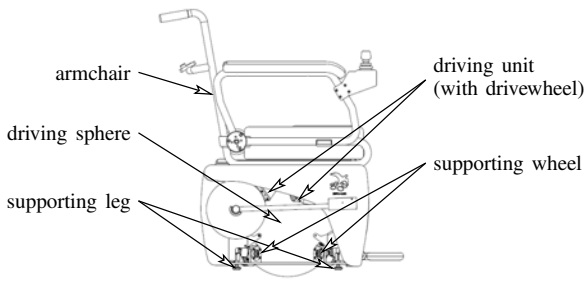


Fig. 3. Side view of the wheelchair; drivewheels set in contact with north hemisphere rotate the sphere.

The power of the servomotors and reduction ratio of the gear train are determined by evaluating total mass of the wheelchair and the user and maximum speed and torque. Other electrical components such as batteries, servo drivers, an embedded controller, and an inertial measurement unit are mounted on the front and back side of the chair.

III. SYSTEM MODELING

To design the stabilizing controller, a full dynamic model of the wheelchair is derived using Lagrange's equations. Assuming that there is no slipping between the sphere and the floor, position of the sphere and attitude angles of the chair are taken as configuration variables. Following the steps shown in [2], linear approximation of the equation of motion is finally obtained:

$$\begin{bmatrix} \overline{\boldsymbol{M}}_{11} & \overline{\boldsymbol{M}}_{12} \\ \overline{\boldsymbol{M}}_{21}^T & \overline{\boldsymbol{M}}_{22} \end{bmatrix} \begin{bmatrix} \ddot{\boldsymbol{q}}_1 \\ \ddot{\boldsymbol{q}}_2 \end{bmatrix} + \begin{bmatrix} \overline{\boldsymbol{H}}_1 \\ \boldsymbol{O}_{3\times 2} \end{bmatrix} \boldsymbol{q}_1 = \begin{bmatrix} \boldsymbol{O}_{2\times 3} \\ \overline{\boldsymbol{B}}_2 \end{bmatrix} \boldsymbol{\tau}_s. \tag{1}$$

$\boldsymbol{q}_1 \in \mathbb{R}^2$ is roll and pitch angles of the chair, $\boldsymbol{q}_2 \in \mathbb{R}^3$ is yaw angle of the chair and the sphere position. $\boldsymbol{\tau}_s \in \mathbb{R}^3$ is driving torque for the sphere. The coefficient matrices in (1) consists of inertial parameters of the wheelchair, which are obtained by mass property report of 3D CAD software. The blocks $\overline{\boldsymbol{M}}_{11} \in \mathbb{R}^{2\times 2}$, $\overline{\boldsymbol{M}}_{22} \in \mathbb{R}^{3\times 3}$, and $\overline{\boldsymbol{B}}_2 \in \mathbb{R}^{3\times 3}$ are nonsingular. Viscous and Coulomb friction of the gear reducer and rolling friction of the driving sphere are identified through experiment; they are used in computing actuator torque command based on (3) shown below.

IV. CONTROLLER DESIGN

The linearlized model (1) is transformed into acceleration input form:

$$\begin{bmatrix} \overline{\boldsymbol{M}}_{11} & \boldsymbol{O}_{2\times 3} \\ \boldsymbol{O}_{3\times 2} & \boldsymbol{I}_3 \end{bmatrix} \begin{bmatrix} \ddot{\boldsymbol{q}}_1 \\ \ddot{\boldsymbol{q}}_2 \end{bmatrix} + \begin{bmatrix} \overline{} & \boldsymbol{M}_{12} \\ & \boldsymbol{I}_3 \end{bmatrix} \boldsymbol{u}, \tag{2}$$

where $\boldsymbol{u} \in \mathbb{R}^3$ is acceleration of the sphere $\ddot{\boldsymbol{q}}_2$. Once the stabilizing controller for (2) is designed, the control input $\boldsymbol{u}$ is calculated based on the state variables of wheelchair. Then the corresponding $\tilde{\boldsymbol{u}} \in \mathbb{R}^3$ with position servoing is determined and finally necessary driving torque of the sphere $\boldsymbol{\tau}_s \in \mathbb{R}^3$ is computed using

$$\boldsymbol{\tau}_s = \overline{\boldsymbol{B}}_2^{-1}(\overline{\boldsymbol{M}}_{22} - \overline{\boldsymbol{M}}_{12}^T \overline{\boldsymbol{M}}_{11}^{-1} \overline{\boldsymbol{M}}_{12})\tilde{\boldsymbol{u}} - \overline{\boldsymbol{B}}_2^{-1} \overline{\boldsymbol{M}}_{12}^T \overline{\boldsymbol{M}}_{11}^{-1} \overline{\boldsymbol{H}}_1 \boldsymbol{q}_1. \tag{3}$$

The minimum norm actuator torque can be decided according to $\boldsymbol{\tau}_s$, based on the kinematics of the sphere and the driving omni wheels. Fig. 4 is the overall controller scheme.

V. EXPERIMENTAL RESULTS

The above controller has been implemented on the embedded controller running RT-Linux and the wheelchair has been successfully stabilized. The performance was evaluated through experiments; Figures 5 are a few of the results, which shows attitude stability of the chair and the climbing ability.

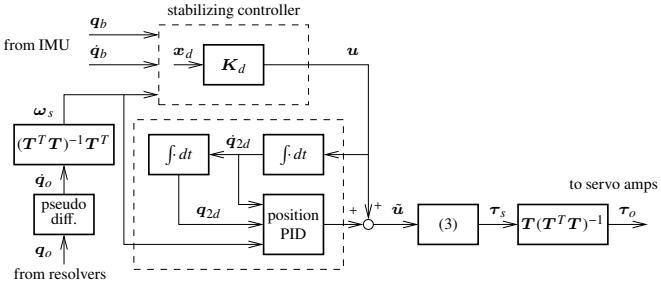
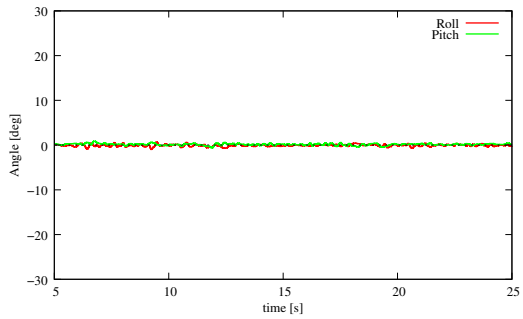
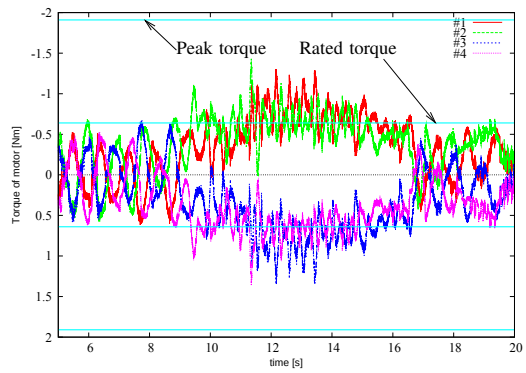


Fig. 4. Controller scheme; $\boldsymbol{q}_o$ is the drivewheels' rotation, $\boldsymbol{q}_b$ is attitude angle of the chair, $\boldsymbol{\omega}_s$ is rotational speed of the sphere, and $\boldsymbol{\tau}_o$ is the drivewheels' torque. $\boldsymbol{T}$ is the transmission matrix from $\boldsymbol{\omega}_s$ to $\dot{\boldsymbol{q}}_o$.



(a) The roll and pitch angle of wheelchair



(b) The torque of four actuators (rising slope of 4.7 degree)

Fig. 5. Experimental results: a) The roll and pitch angle of wheelchair tracking a 2 meters square trajectory; the wheelchair is kept upright. b) The torque of four actuators while the wheelchair is climbing a rising slope of 4.7 degree; they remain at the rated value.

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Development of Ankle Rehabilitation Device which can Mechanically Avoid Undesirable Load*

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***Abstract*— The authors introduce the development of ankle rehabilitation device which has a specially designed spatial mechanism to avoid undesirable loading force and torque while transmitting only the desired driving torque to cause flexion/extension of ankle joint. Since the authors came up with the mechanism composition, four prototypes including an initial engineering prototype have been fabricated, and recently the latest prototype is going to be used in clinical test in a hospital. In this paper, the history of the prototype fabrication while improving the design implementation is introduced.**

I. Introduction

In order to establish sustainable highly-aged society, there is a social requirement in many developed countries that R&D agencies such as academic universities, hospitals and private technology companies have to pay great effort to make innovations on various engineering solutions. One of the most important requirements for such effort is maintaining one's walking function as long as possible since it is a very important function for people's daily life. One of the most popular topic on this field is walking assistance technology such as exoskeletons which is recently gathering a lot of attentions, but the authors has taken more basic and inexpensive approach, a specially designed rehabilitation device for human's ankle joint. In this paper, the concept and entire history of the R&D progress will be explained.

II. Concept of the Ankle Rehabilitation Device

In the year 2012, a research group of Mechanical Systems Design Lab in Tokyo Institute of Technology (TokyoTech) initially came up with a concept of a human's ankle joint rehabilitation device[1],[2]. The key feature of the concept was that the device had a specially designed mechanism which avoids undesirable loading force and torque while only transmitting essential driving torque around the axis of rotation of the joint. Since human's ankle joint is composed of three bone pieces (talus, fibula and tibia) tied by ligaments, its center of rotation continuously moves while the ankle joint rotates. In order to achieve the above-mentioned function, a spatial coupling mechanism which was derived as an extended Oldham coupling mechanism was included as shown in Fig.1. The illustrated mechanism in Fig.1 is having not only the spatial coupling mechanism ($CDEFO_a$) but also a mechanical regulation of range of motion (ROM) which was implemented as a four-bar linkage closed-loop mechanism (OABC) attached to the driving link CD. This mechanism becomes 1-DOF only when the user's lower limb was taken into account (without user's lower limb, the mechanism has 6-DOFs). This means that the device is functional only when it is attached to user's lower limb, and it can adapt arbitrary user's physique and rotation axis' duration while in action without any computer control or sensory information.

III. Progress of the Prototyping under a Joint R&D

a) First engineering model

According to the design concept shown in Fig.1, an engineering model shown in Fig. 2 was fabricated. By using this model, a dummy-leg was driven while measuring the load by a force/torque sensor. From obtained results, effectiveness of the spatial coupling mechanism for suppressing undesirable load and adapting to arbitrary axis movement of the target joint was demonstrated. On the other hand, it was also found that friction force especially on cylindrical joints caused undesirable load significantly, and it was thus necessary to make the device as lightweight as possible while implementing gravity compensation function. In order to compensate the load due to the gravity, and also to give a desired load on the ankle joint, adjustment by using springs attached to certain joints was also discussed.

b) First wearable prototype (Ver. 1)

After that, a first wearable prototype shown in Fig.3 was fabricated[3]. This prototype had a ROM control function based on the four-bar mechanism as already mentioned. In addition, it was very lightweight because its frame and linkages were made of Ultra High Molecular Weight Poly Ethylene (U-PE), and the electric motor could be driven by a 9V dry-battery.

c) Prototype for commercial production (Ver. 2)

After doing experiments with the first wearable prototype, the TokyoTech group has started to discuss with medical people who have practical knowledge on rehabilitation in the year 2015. At this time, a conceptual representation had been improved as illustrated in Fig.4. In this figure, lengths of the two cylindrical joints were enlarged to reduce friction caused by the gravitational load, and both side of the C-joint's shaft were held to prevent the joint from falling off while a user is wearing the device. It was also found that the way of fastening

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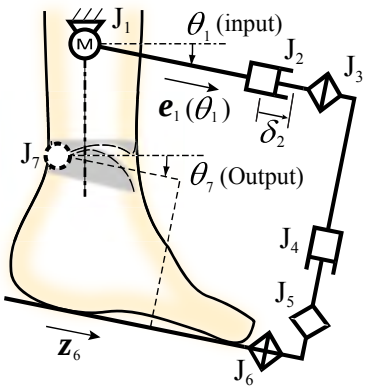

Fig.1 Composition of a spatial mechanism for ankle joint rehabilitation.

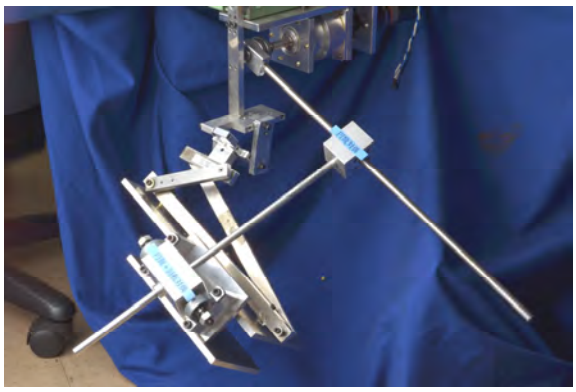

Fig. 2 An engineering prototype for concept demonstration. (Mar. 2012)

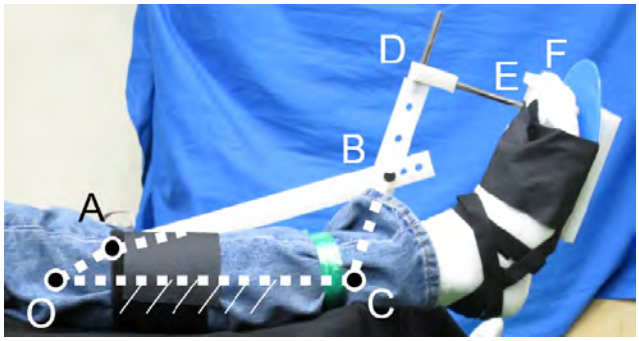

Fig. 3 Prototype ver. 1. (Jul. 2014)

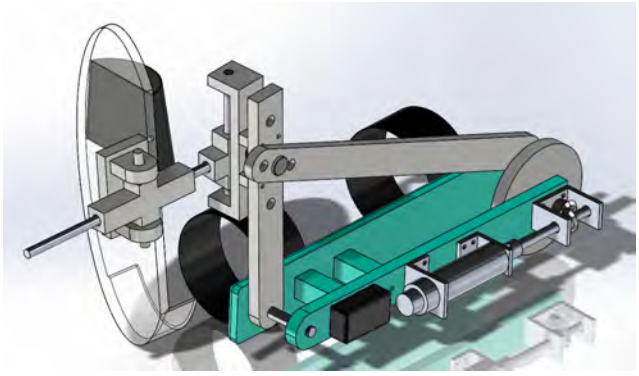

Fig. 4 Draft proposal (3D-CAD) of revised design of ver. 1. (Jul. 2015)

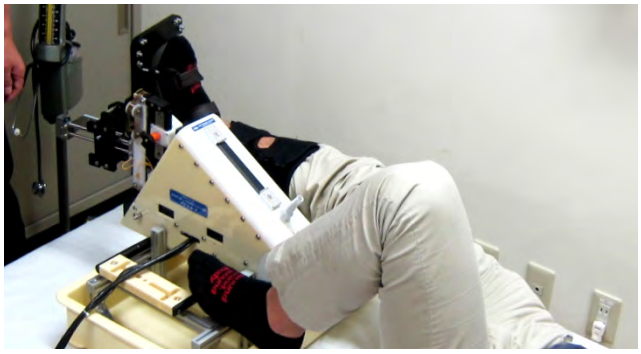

Fig. 5 Prototype ver. 2. (Sep. 2016)

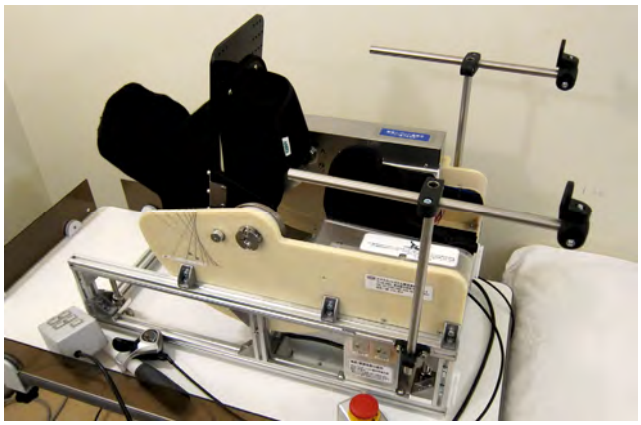

Fig. 6 Prototype ver. 3 with double-sided support feature. (Jan. 2017)

the device's body to a user's lower limb should be improved to prevent undesirable deflection while the device is in motion.

At the beginning of this research, the purpose of the device was originally assumed to be ROM extension and motor function recovery after surgical operations such as suturing of Achilles tendon. In addition to this, other potential applications such as recovery of muscle fatigue, thrombus prevention treatment of deep vein while patients are kept in unconscious during surgical operations, or rehabilitation of disabled lower limb function for patients of cerebral stroke, were found out.

In order to develop a useful rehabilitation device for above possible use cases, a joint research and development group has been established by TokyoTech, Tokyo Rosai Hospital and Abe Techno System, Co. in the year 2016 to manufacture a commercially available product. Abe Techno System is a company having a lot of experience for prototyping and mass-production of various commercial products and has had initiative for development of the latest two prototypes shown below. As a first product of this collaboration, a prototype for commercial production shown in Fig. 5 was fabricated. In this prototype, many practical implementation such as compensation of gravitational load by using springs, compaction of slider components, insertion of torque limiter in-between a driving motor's reduction gear shaft and user's ankle joint to prevent too large torque exertion, and introduction of micro-controller to achieve flexible rehabilitation tasks. This prototype was wore by a physio

therapist in the R&D group. It was said that ROM and strength of the exerted torque were not harmful.

d) Second prototype for commercial production (Ver. 3)

In order to solve some issues found in the ver. 2 prototype and to proceed clinical tests, a latest prototype, ver. 3, was fabricated. Modification of mechanism has done to reduce friction on each joint and undesirable load on user's foot. Currently, the joint R&D group is continuing evaluation of the device and preparation for clinical tests with general users at Tokyo Rosai Hospital.

IV. Conclusion and Future Work

In this paper, the concept and R&D history of the ankle joint rehabilitation device was introduced. It can be said that the joint discussion and R&D project with medical people and private developing company are fruitful, since many improvements while making several prototypes and evaluating them in a hospital have been done in such a short time. In near future, effect of the device will be evaluated by using NIRS and fMRI equipment.

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The importance of transparency in rehabilitation robot design and its evaluation

Justin Fong, Vincent Crocher, Iven Mareels, Ying Tan and Denny Oetomo

I. INTRODUCTION

Motor recovery from neurological injuries is driven by intensive therapy involving repetitions of goal-orientated movements. To assist this, a number of robotic devices designed for use in the rehabilitation of the upper extremity for neurologically impaired patients have been developed over the past 20 years. Such devices mechanically interact with the patient whilst they attempt to perform motor actions, either assisting or challenging the patients in a structured way, aimed at accelerating and/or furthering their recovery. One characteristic of rehabilitation of neurologically impaired subjects — or neurorehabilitation — is that generally the human physical system, the limb at the biomechanical level, is intact and that it is the sensory motor loop that needs to be restored. This requires that rehabilitation encourages active patient participation as much as possible, as early as possible. As such, one critical characteristic of rehabilitation robots is their transparency [1], such that this intent and low levels physical activity of the patient can be detected.

Backdrivability of a mechanical system or robotic system is how much of its own mechanical dynamics is presented at the end-effector, and is significantly affected by the system's friction and inertia. It is a contributing factor to the dynamic transparency of the device. In terms of neurorehabilitation, transparency directly translates to the ability of the system to feel and react to the patient movements or even movement attempts. This thus allows the system to 1) implement 'assist-as-needed' control strategies [2]; 2) reduce over time as the patient recovers; and 3) not prevent or contradict any correct movement of the patient. It is important not to understate the significance of this feature — humans are sensitive to extremely low level of forces (as low as 1 Nm at a joint [3]) and stroke survivors, the target population of these devices, may exhibit very little movement capabilities which should be detected.

This particularity makes rehabilitation robots very different in their design from standard industrial robotic systems where force capability, accuracy and repeatability are usually the main requirements.

As for many aspects of rehabilitation robotics, it is difficult to translate an evaluation of the transparency from the engineering to the clinical space. Indeed, different methods of evaluation exist and range from purely engineering approaches to those which are more human-centred. One traditional method is to use a force and torque sensor to measure the forces applied at the end-effector when a given motion is performed [4]. In this case, the smaller the magnitude of measured force and torque, the more transparent the system can be considered. Alternatively, within the context of rehabilitation of the upper limb, transparency can also be evaluated by having human subjects perform reaching motions while they are attached and not attached to the rehabilitation robot. In an ideal case, the trajectories for the same intended motion would be identical — *i.e.* the robotic device does not affect the movements of the subjects.

We present here the method we used to evaluate a 3D manipulanda we developed for upper-limb rehabilitation. This is inspired from [5] and already used on the ArmeoPower, a commercially available exoskeleton, which provides a first point of comparison.

II. EMU: A 3D MANIPULANDUM

Within this work, a transparency evaluation technique is applied to a 3D manipulandum developed for upper-limb rehabilitation. This device has been designed as a trade-off between planar manipulanda and complex exoskeletons and have the following characteristics:

- Large achievable range of motion in three dimensions.
- Ability to (indirectly) correct subject postural behaviour through monitoring of arm posture.
- Possibility to interact with real world objects — to avoid the use of an extra virtual mapping (hand-to-screen).
- A wide range of dynamic bandwidth, from a very transparent setting to a reasonable amount of stiffness.

The EMU has 6DOF in end-effector movement, where only the first 3DOF (associated with the translation movement of the end-effector) are actuated. The first axis is rotational about a vertical axis. The second and third actuate a 4-bar linkage arrangement in a vertical plane positioned by the first axis — see Figure 1. Inertia is limited due to the positionining of the motors at the base of the manipulandum, and backdrivabilitiy is preserved through actuation of these axes through a capstan transmission. An unactuated ball-joint is constructed for the end-effector, which is connected to the patient's wrist utilising a splint such that the center of the wrist corresponds to the robot end-effector point.

On the controller side, the software is designed in a hierarchical manner, with the open-loop gravity and friction compensations and an impedance controller [6] running at 10kHz on an FPGA and higher level controllers running at 1kHz on a RT controller.

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Fig. 1. The EMU prototype with a subject.

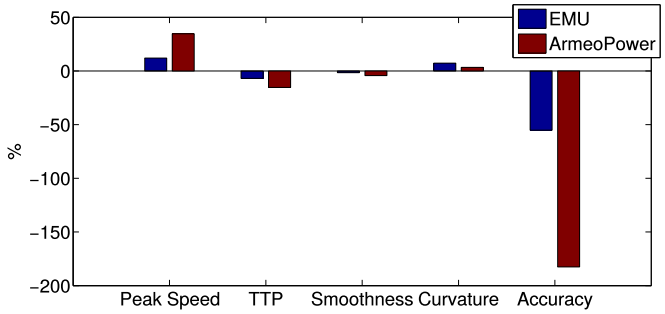

Fig. 2. Comparison of change in metrics (percentage) when performed within EMU, and when performed within the ArmeoPower

III. Transparency Evaluation

The approach used here measures the transparency not in terms of engineering physical measures but rather observes the effects of the robot on human movements during a nominal rehabilitation exercise. This provides an application-based metric, which is easily explained to therapists.

The method is experimental and is described in detail in [7], where it was applied to the ArmeoPower exoskeleton. Within the present experiment, five healthy subjects were asked to reach to virtual targets in two conditions: (1) out of the robot and (2) in the robot set to its transparent mode (compensation of its own weight, and friction). Their arm movements were monitored using magnetic positioning sensors (3D Guidance trakSTAR, Ascension Corp). The position of the wrist was mapped to a virtual cursor, and subjects were asked to reach from a fixed starting position to six different virtual targets representing movements to different areas of the workspace infront of the subject.

The impact of the robotic device on subject performance was measured using five metrics dependent on wrist position only and chosen for their relevance to rehabilitation [8]:

1) Peak Speed: the largest speed of the movement.
2) Time of Peak Speed: the time to reach the peak speed;
3) Smoothness: the Spectral Arc Length (SAL) Smoothness as defined in [9];
4) Curvature: the integral of the distance of the reaching trajectory from a straight line;
5) Accuracy: the distance to the target at $t = 1$s.

IV. Results and Discussion

Movements made within the EMU were affected compared to those made outside it. However, these effects are small, with Peak Speed, Time to Peak Speed, Smoothness and Curvature affected by less than 15%. Accuracy is affected more significantly (a 50% decrease). However, the absolute change is approximately $3mm$ in magnitude. Thus, although the subjects were aware of the EMU, its effect was minimal.

Figure 2 illustrates the percentage change from *in robot* to *out of the robot* between the two different robotic devices — the ArmeoPower, and the EMU. In this comparison, it can be seen that the changes in the metrics introduced by the EMU are two to four times lower than the ArmeoPower. Such results are not unexpected, due to the difference in design priorities of the ArmeoPower (and a more targeted patient segment). This leads to a different structure, with the ArmeoPower being a full exoskeleton, thus being attached to the arm at multiple locations at which force can be imparted on the subject. Additionally, the ArmeoPower's serial structure naturally leads to a heavier system and thus more inertia which must be compensated for.

V. Conclusions

The measurement of transparency can be achieved by investigating its effect on human movements. In this work, a experimental method was used, demonstrating a robot's ability to preserve movement intention of the subjects and to not significantly affect their movement patterns. This method of evaluation has the advantage of being applied directly in the space of interest, and reflects the potential in-homogeneity of the robotic device (resulting from the device's structure) and its consequence on the human.

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Individualised and adaptive upper limb rehabilitation with industrial robot using dynamic movement primitives

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***Abstract*—Stroke is a leading cause of serious long-term disability. Post-stroke rehabilitation is a demanding task for the patient and a costly challenge for both society and healthcare systems. We present a novel approach for training of upper extremities after a stroke by utilising an industrial robotic arm and dynamic movement primitives (DMPs) with force feedback. We show how pre-recorded and learned DMPs can act as basis exercises, that can be modified into individualized and adaptive rehabilitation exercises that fit with the patient's physical properties and impairments. We conclude that our novel approach allows for easy and flexible set-up of rehabilitation exercises and has the potential to provide the therapists and patients much easier interaction with such complex technology.**

I. Introduction

Stroke is a leading cause of serious long-term disability and reduces mobility in more than half of stroke survivors age 65 and over [1]. Rehabilitation after a stroke is a huge and demanding task for the patient and a costly challenge for both society and healthcare systems [2], [3], [4].

Several studies have shown that after a stroke, increased amounts of task repetition causes cortical changes and functional improvement [5], [6], thus, letting the brain re-learn to control the paralysed muscle groups [7], [8], [9].

Depending on the nature of the functional impairment, the patient is often unable to perform exercises without the help and physical support of a therapist. This decreases amount of task repetitions practically possible. In the Patient@home project we aim at making a robotic system which is able to rehabilitate the upper extremities. The system must be flexible enough to support and be adaptive to the many different impairments that people suffer after a stroke. One solution to this challenge is to provide the therapist with a standard set of rehabilitation exercises that can be individualized for each patient. This will make it less time consuming for the therapist to set up a specific exercise - e.g. seen in contrast to having to record each specific exercise for each patient.

Many other projects investigated the use of electromechanical and robotic devices for rehabilitation [10], [11], [12], however, those systems usually do not use feedback or modification of training movements during the training process and, therefore cannot adapt to particular needs of a specific person. Currently, within the field of robotics, *dynamic movement primitives* (DMPs, [13], [14]) are some of the most common choices for generation of robot motions, due to their attractive features such as generalization to new start-/end-points, robustness to perturbations, and ability to modify trajectories on-line by ways of learning and/or sensory feedback. Recently, the usage of movement primitives in rehabilitation and physiotherapy has been suggested by N. Hogan and D. Sternad [15].

Here, we present a novel approach for training of upper extremities by utilising an industrial robotic arm and DMPs with force feedback [16], where a set of pre-recorded and learned DMPs act as training exercises for individualized and adaptive rehabilitation exercises that fit with the patient's physical properties (e.g., short arms vs. long arms) and impairment.

II. Methods and Materials

Our setup consists of a UR5 industrial robot from Universal Robots, A Robotiq force-torque sensor, a PC for running the developed java control software and a tablet for running the web-application. Our robot control software is built upon the RobWork framework [17], which is a collection of C++ libraries for simulation and control of robot systems. We mounted a handle on the force-torque sensor allowing the user to grab hold of the robot. Additionally, gloves with Velcro and wrist support are used to support patients that do not have enough grip strength to hold on to the handle.

The research leading to these results has received funding from the Danish agency for Science, Technology and Innovation Project Patient@home under grant agreement no. 11-116201 and the European Communitys Horizon 2020 FETPROACT-2016 Framework Program under grant agreement no. 732266, Plan4Act.

First, a basic set of training trajectories (exercises) is recorded using the UR5 robot-arm. This set is based on standard exercises used at the neurorehabilitation department at Odense University Hospital, Denmark, and include supination and pronation of the forearm and wrist, as well as flexion and extension of shoulder and elbow as shown in Figure 1. Usually, these exercises are done along a table surface, such that the arm's weight is supported by the table. In this project, the robot replaces some of the support provided by the table, giving physical support to overcome both gravity and friction. The basic set of movement primitives chosen are 1) straight movement (translation) of the arm, 2) a rotation of the wrist, and 3) curved movement.

(a) Flexion extension (b) Pro-/supination (c) Rotation

Fig. 1. Images of three basic exercises.

Afterwards, these trajectories are encoded and learned by using a version of DMPs for interaction learning as proposed by [16]. We use force feedback in order to modify and adapt trajectories on-line. Note, that in our case a reaction to forces is not learned, but predefined manually.

Finally, learned DPMs are used for individualised training by

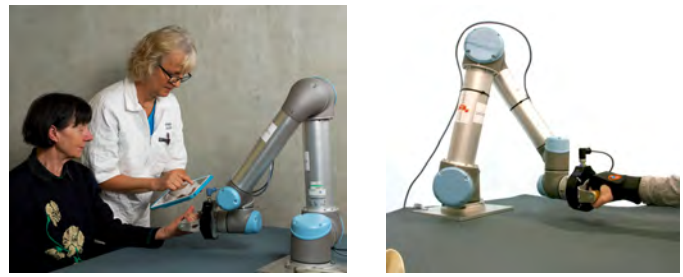
- changing start-/end- points of given movement primitive;
- combining primitives, e.g., translation of an arm with a rotation of a wrist;
- altering velocity profile of the movement;
- altering position profile of the movement based on robot-patient interaction by utilising force feedback.

III. Results

What is presented here is a series of results, that each count as foundational building-blocks in our overall approach to making a flexible and adaptive rehabilitation partner for individualised rehabilitation. Figure 2 a) shows how therapist and "patient" together define basic exercises, which are then recorded through the browser interface where from the therapist and patient can initiate recording and execution of training exercises. This interface is also used to allow the patient to later train using adapted / adapting versions of the originally recorded exercise.

A. Changing of start- and end-points

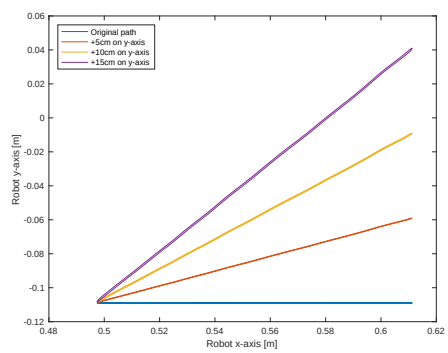
Changing start- and endpoints is an important necessity, when configuring exercises for different types of patients and impairments. First, patients rarely have identical physical properties, which means range of motion is different. Second, different types of impairments means different types of patient



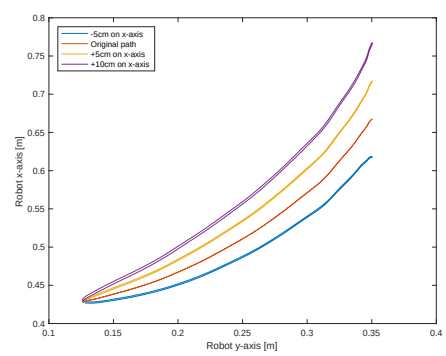
(a) Recording exercise (b) Patient training

Fig. 2. Setup of the robotic-arm for training.

arrangement during training - e.g. some patients can sit, while others have to be laying down. Therefore the physical placement of the robot arm in relation to the patient is almost never the same. Figure 3 a) shows how we are able to modify the standard flexion extension exercise displayed in Figure 1 a) displacing the endpoint along the y-axis, which would for instance be necessary if a patient is seated at another angle in relation to the robot. Figure 3 b) is displaying the same kind of scaling of the rotation exercise also shown in Figure 1 c).



(a) Flexion extension scaled



(b) Rotation scaled

Fig. 3. Generation of new movement trajectories by changing end-points.

B. Combining primitives

Often, patients train different types of movement separately before combining them. This could for instance be the flexion

extension exercise displayed in Figure 1(a) and the pro-/supination exercise shown in Figure 1(b). Merging these two would lead to a forward translational movement while rotating the wrist, while chaining them would lead to an exercise with these two exercises performed in succession.

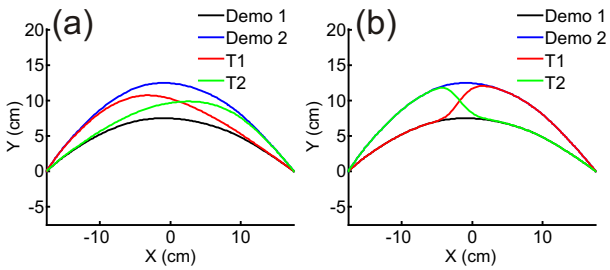


Fig. 4. Simulation results for merging of movement primitives: (a) a gradual transition from Demo 2 to Demo 1 (trajectory T1) or from Demo 1 to Demo 2 (trajectory T2), and (b) an abrupt transition. Demo 1/Demo 2 - learnt demonstrated trajectories, T1/T2 - generated new trajectories.

Simulation examples of generation of new trajectories by merging two basis motions (Demo 1 and Demo 2) are shown in Fig. 4. In panel (a) we show trajectories T1 and T2 which emerge from gradually shifting from Demo 1 to Demo 2 (trajectory T2) or vice versa (trajectory T1), whereas in panel (b) we show emerged trajectories from an abrupt transition. Also, other trajectories can be generated by merging trajectories with different weighing functions.

C. Altering velocity profile

Changing the velocity of an exercise is also key in rehabilitation. Usually, right after a stroke, patients can only move their impaired limbs very slowly - if at all. Building up speed goes hand in hand with building up strength or regaining muscle control, and therefore a robot training partner has to be able to detect and support this progress. Also velocity control is a necessary step in making natural force feedback using the robot force sensor. Figure 5 shows how we change the speed of the flexion extension exercise (Figure 1 a)). The red curve displays the robot back and forward motion at constant speed, while the blue curve shows the motion at varying speed.

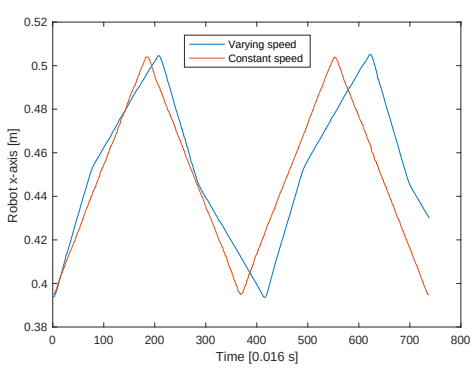


Fig. 5. Generation of trajectories with a varying speed.

D. Altering position profile

During rehabilitation patients often increase their range of motion - e.g. the number of degrees you can rotate around your shoulder joint. Increased range of motion is, like increased speed, a sign of progress in the rehabilitation process, and therefore the robot has to be able to acknowledge this progress - e.g. through its force sensing and adapt by increasing the range of motion accordingly. Using the robot force/torque sensor output we succeeded in first varying the speed of any given exercise with regards to the force applied - allowing the user to only follow the defined DMP path. Figure 6 shows how this is done in simulation. Experiments still have to be conducted on the real robot.

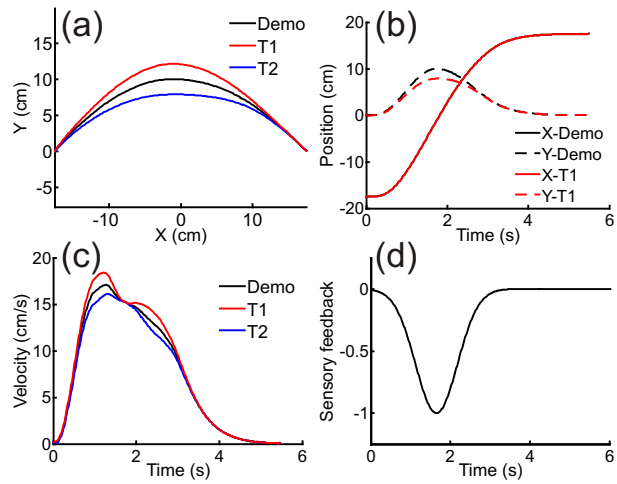


Fig. 6. Simulation results for altering position profile with a sensory feedback (e.g., force): (a) position profiles, (b) X and Y components for trajectories Demo and T1 (see panel [a]; X and Y components for T2 are not shown), (c) squared velocity for both X and Y components, and (d) sensory feedback for the T1 case. T2 was generated using sensory feedback of similar shape as in panel (d) but with a positive sign, i.e., varying from 0 to 1. Demo - learnt demonstrated trajectory, T1/T2 - generated new trajectories using sensory feedback.

We are using this method, so that when the system experiences increasing force exerted at the endpoints of e.g. the flexion extension exercise we can dynamically change the endpoints of the exercise and thereby adapt to the user intentions and/or increased range of motion.

IV. Conclusion

In this study we presented a novel approach for automated training of upper extremities by utilising industrial robot arm and dynamic movement primitives (DMPs) with force feedback. We demonstrated that our approach enables individualised and adaptive training for persons with different physical properties and impairments. Although some parts of this are mainly simulations and still needs to be extensively tested, we see the use of DMPs as a step in the right direction in order to allow for easy and flexible set-up of rehabilitation exercises allowing the therapists and patients much easier control over and interaction with such complex technology.

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Natural Kinesthetic Interaction and Social Relations Between Training-Robots and Their Users

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***Abstract*—Our transdisciplinary work on robotic training partners for rehabilitative and preventive health care training combines health, social, engineering, and computer sciences into a robotic training technology that facilitates natural and intuitive interaction between the human user and the robot training partner. We introduce our work and analyses of forceful high-bandwidth kinesthetic human robot interaction (HRI) in rehabilitative and preventive health care training situations, including both healthy and disabled subjects. The HRI is considered in the context of Social Embodied Interaction (SEI) analysis. We demonstrate how kinesthetic HRI is quite natural and intuitive. We consider unobtrusive interaction, designed to integrate into the Human Participants (HP) embodied interaction with the perceived world — seeing but not noticing the Robot Participant (RP). We also demonstrate how the kinesthetic HRI can be designed to commandeer attention — causing the HP to notice and socially respond to the RP, with bodily force and expressive body language, in an intuitive way. We conclude with a discussion of applications for these types of interactions in the rehabilitative and preventive health care training of patients and other users.**

I. Introduction

Using robotic devices as training partners has enabled us to study human-robot interaction (HRI) where information is exchanged primarily through kinesthetics, rather than audio, visual, or tactile means. Our *RoboTrainers* enable us to push, pull and overpower a human participant (HP), with dynamics that exceed the response time of the human neuromuscular system. This allow us to perform HRI experiments, where the Robot Participant (RP) appears completely unobtrusive, allowing the HP to interact bodily with a world that is strongly augmented by the RP without focusing attention toward the RP itself. Alternatively, we can create HRI experiments with an obtrusive RP, where the HP strongly directs attention toward the RP with expressive body language and strong bodily interaction. We can thus study the ”language” of kinesthaetic HRI, and use the results to create specialized physical training programs designed to optimize the effort expended during training and its effect. Currently, we are addressing rehabilitative training for orthopaedic and neurological patients, as well as preventive training for astronauts in weightless conditions.

II. Related work

Research and development of training robots is abundant, with several products already maturely on the market. Although the point of training robots is physical human/robot interaction, interaction mostly appear to emerge as a result of optimizing a training program, not as a separate topic of interest. Robot-assisted training devices roughly fall into four categories:

a) Robotized training machines: The normal passive loads from weights, springs, inertia or friction is substituted by computer-controlled actuators with sensor feedback. These machines have a high level of mechanical support for the body and limbs, at the cost of limited kinematic freedom. Typically, the spatial motion path is defined by the mechanical configuration, while velocity and force are controlled by the computer. Such machines are becoming widely available for rehabilitative and fitness training through commercial products, such as the “gym80” portfolio [1], and has previously been available for researchers as specialized or custom-built equipment.

b) Linked robots: Utilize the high kinematic freedom of serial link mechanisms to let both the spatial motion path, velocity and force be computer controlled. They typically only interface to the user at a single point, through a handle or splint. Typical examples like MIT’s Mc-Manus [2] remain too specialized to enter high volume production, and hence too expensive to appear outside research and hospital use. With the advent of human-safe mass produced robot arms, it has become an option to use commercial off the shelf technology (COTS) as a base for linked training robots in research [3], [4], [5] as well as commercially [6]. This type of robot will probably become feasible for ordinary hospitals and well-off community training centers, but they will hardly become feasible for personal training of ordinary people at home.

c) Exoskeletons: Are highly specialized articulated robots with kinematics that match the kinematic of the human body, making them “wearable” [7]. Exoskeletons combine the

kinematic freedom of linked robots with the rigid support of training machines, making them the ideal training robot from a technical perspective, although their complexity and price still inhibit widespread use.

d) Wire robots: Consist of one or more computer controlled winches that control or assist spatial motion through wires or ropes with sensor feedback. More advanced systems, such as STRING-MAN [8] and NeReBot [9], use multiple synchronized winches in a parallel kinematic configuration, to control the motion in multiple dimensions, but the principle can be reduced to a single wire, like Bioness Vector[10].

III. Our training robots

We have developed a wide range of experimental training robots, that work well as experimental platforms for exploring physical interaction, as well as studying the effects of training.

A. Robotrainer-classic

Combining a classic biceps/triceps "Curl machine" with force sensors, a large motor, and an advanced low-latency control unit, has given us a very powerful experimental platform: RoboTrainer-Classic (RTC). It can overpower any human participant, but also provide gentle force-assist for partially paralyzed participants.

Max Force	$1000[N]$
Force resolution	$0.1[N]$
Max Velocity	$200[deg/s]$
Range	$120[deg]$
Resolution	$0.1[deg]$
Control frequency	$800[kHz]$

Fig. 1. RoboTrainer-Classic - main features

B. Robotrainer-Light

Combining the same control unit, with a small wire-pulling motor and a force sensor has resulted in a versatile research and demonstration platform: RoboTrainer-Light (RTL). A detailed description is available in [11], but the main features of the current version are summarized as:

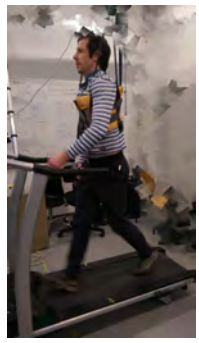
Max Force	$250[N]$
Force resolution	$0.1[N]$
Max Velocity	$1.5[m/s]$
Wire range	$3[m]$
Wire resolution	$1.5[mm]$
Control frequency	$800[kHz]$

Fig. 2. RoboTrainer-Light - main features

The low-level controller is implemented on a Spartan-6 FPGA [12], that exchanges set-points, measurements and control parameters with a PC or similar computer through Unity-Link [13]. Training/interaction programs running on the PC periodically sample measurements from the robot, and adjust controller set-points or control parameters according to the required training/interaction.

C. Robotrainer-Lift

Updating motor, gearbox and sensors, but maintaining the control system, changes RoboTrainer-Light into RoboTrainer-Lift. This system is able to lift an entire person, but operate at very reduced velocities for safety reasons.



Max Force	$2000[N]$
Force resolution	$0.5[N]$
Max Velocity	$0.1[m/s]$
Wire range	$2[m]$
Resolution	$100[\mu m]$
Control frequency	$800[kHz]$

Fig. 3. RoboTrainer-Lift - main features

D. Control unit

The low-level control unit is based on the TOS-NET/Unity-Link Field Programmable Gate Array (FPGA) based architecture, developed previously [12] [13]. The architecture support high performance rapid prototyping of complex electronic interface circuits, and allow us to utilize the full potential of any type of electric motor or sensor we choose to integrate. The parallelism and performance offered by FPGA's ensure that the control unit is not limited by the threading, latency and I/O buffer issues that are common in conventional embedded controllers. All controller components operate in parallel on the hardware level, with possible update rates ranging into GHz, although the current update range of $800kHz$ is more than sufficient for the systems described here.

Figure 4 show how the major components of the mechanical platform, the low level control unit responsible for velocity, force and position control, and A PC-type computer running the high level interaction programs. As an addition to previous work, we have integrated the Robot Operating System (ROS) in the "transport" layer of the system, to allow easy programming of systems with multiple robot modules.

The controller (CTRL) regulate the dynamic state of position, velocity and force, based on setpoints from the PC and measurements from sensors. This *state* essentially define the momentary element of the physical interaction between human and robot participant. The high bandwidth of the controller ensure that the robot react to state changes much faster than the human can initiate or perceive them.

Programs running on the PC can sample the current dynamic state, change the state set point and/or the control parameters, in order to react to or initiate changes in the momentary state of interaction. The current technology enable the PC to sample and update state parameters, with a latency of approximately $1ms$, allowing interaction programs to obtain sample rates of up to $1000Hz$.

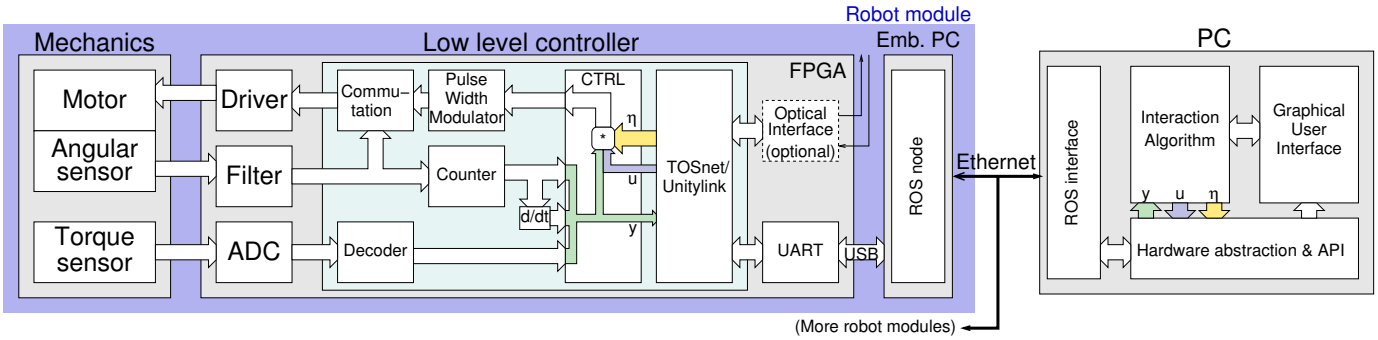


Fig. 4. TOS-NET/Unity-Link based control system

IV. Social Interaction Methodology

The interaction between human participant (HP) and robot participant (RP) is analyzed in the context of *Social Embodied Interaction* (SEI). How people understand social actions in the context of an environment, place, position, movement and time. The meaning of an action for a HP is worked out by the action's position in an action-sequence, how the actions are performed by body motion and how people coordinate these bodily actions down to minute details. A handshake is a typical example of such a sequence, containing the steps: Invitation, initiation, shaking, invitation to end, and ending.

Our early experiments were focused on training physiology, where the robots were programmed to optimize specific aspects of physical training. The emerging HRI was later studied in the SEI context, and inspired us to design experiments with the primary purpose of studying Human/Robot SEI.

V. Unobtrusive robot participant

The high control bandwidth of our robots allows them to measure and adjust their force and motion faster than the human participant (HP) is able to sense and react. We have successfully used this feature to simulate changes to gravity that is accepted by the HP's as a natural altered condition of the interaction between the physical world and their own body. In the experiments described below, the HP interact bodily with the RP, focusing their attention on the training activity, not the RP, although the manifestation of the physical world is strongly augmented by the interaction with the RP.

A. Removed gravity in nerve damage rehabilitation

Traffic victim Jesper Kiersgaard suffers from peripheral nerve damage in his left arm reducing biceps force to $10N$ — too little to overcome gravity. RoboTrainer-Classic is programmed to match the gravitational pull on the arm with an exact counterforce during the entire curl motion, creating an illusion of weightlessness, combined with an amount of dynamic friction. The system allowed Jesper to perform curls with a velocity of $40[deg/s]$. While Jesper is only lifting 40% of his arms weight on the affected side, he interacts bodily with the RP as though he was pulling a normal 20kg weight with a healthy arm, causing him to sweat profusely. During the exercise, he focuses entirely on his arm exercise. In spite of seeing and sitting on the robot, he does not notice it.

(a) Nerve damage rehab (b) Stroke rehab

(c) Weightlessness (d) Walking rehab

Fig. 5. Experiments with augmented gravity

B. Removed gravity in stroke rehabilitation

A RoboTrainer-Light (RTL) is used to counter the gravity acting on the left arm of an elderly stroke victim, allowing the patient to reach markers placed at head-height with her own effort. Once attached to the RTL, the patient was able to move her arm on her own accord. During exercises, the HP ignored the RP, focusing on her arm/hand, markers indicating desired hand positions, and the interaction with the assisting therapist.

C. Simulated gravity in weightless environment

In a partnership devoted to Space Training Technology [14], two RoboTrainer-Light's (RTL) are used with pulleys and a harness, to simulate gravity in an otherwise gravity-free underwater environment. The simulated gravity allows the HP to perform normal strength training exercises as lunges, squats, crunches and push-ups — virtually as they would in normal gravity. The HP's focus their attention on the bodily interaction with the unfamiliar environment, never focusing on the equipment used to facilitate the interaction. Afterward, all HP's referred to experiencing "gravity" — not interacting with robots.

D. Reduced gravity during walk rehabilitation

In a partnership devoted to hospital training a RoboTrainer-Lift was used with a harness, to reduce the gravitational stress on hips and legs during walking on a treadmill. The HP's focus is on the walking and the rather unpleasant pressure created by the less than optimal harness. When asked to abstract from the unpleasantness, HP's describe the experience as: *"Reduced gravity"* rather than *"interacting with a machine"*

VI. Obtrusive robot participant

When RoboTrainer-Light (RTL) is programmed for interactions that do not mimic natural phenomena, the HP relates to the Robot as a sentient, noticable participant, seeking eye contact and using body language appropriate for human/human interaction. This is interesting, as no effort have been made to make RTL human like. It is clearly machine-like, resembling a winch. Furthermore, the HP's are fully aware that the "winch" is controlled by a human operator through a PC, but they do not direct attention toward the operator during their interaction with the RP.

(a) Inverse spring (b) "Fishing" action sequence

Fig. 6. Experiments with obtrusive robot participant

A. Inverse spring

A normal spring will yield when pulled by a human, following Hooks law and become elongated proportional to the pulling force. The RTL was programmed to act as an inverse spring — becoming shorter when pulled. All tested HP's: 1) sought eye contact with the RP; 2) used expressive body language toward the RP, and; 3) pulled very hard, causing the "spring" to become even shorter. In cases where RTL's $250N$ maximum is able to overpower the HP, the HP's body language typically express agitation, even anger toward the RP. Stronger equipment is necessary to expand this test to HP with above average strength.

B. "Fishing" action sequence

In order to study RP initiated action sequences, we have created a RTL interaction program with a simple sequence [15]

1) **Invitation:** RTL will move it's wire up and down in a sinusoid motion, resembling the action of a fishing hook.
2) **Initiation:** When an HP grasps the wire, RTL senses the added force, and changes to a state where it pulls with a moderately high force of $60N$
3) **Moving:** RTL will yield if the HP increases the force and reel in the wire if the HP reduces the force. Thus, the RP and HP will be locked in a gentle $60N$ tug-o-war for a period of time.
4) **Invitation to end:** After a random amount of time, RTL will sharply reduce its pull to a low force of $20N$
5) **Recommencing:** If the HP increases the force to more than $60N$, or pulls the wire longer than $1.5m$, the RTL will sharply increase the force back to $60N$, and go back to the 'Moving' state.
6) **Ending:** If the HP relaxes the force on the wire to less than $5N$ for more than $2s$, the RTL will immediately change back to 'Invitation' mode.

The Fishing sequence was tested on a small group of engineering students who were filmed and interviewed. It was also tested on a larger group during a University Open House event, but without film and interviews. These experiments gave the following results:

- All HP's accepted, understood and interacted through the "Invitation", "Initiation" and "Moving" states.
- All HP's established eye contact with RTL when state changes occurred
- $2/3$ of the HP's accepted the 'Invitation to end'
- $1/3$ of the HP's continued the interaction until the operator asked for termination after a preset 2 minutes.

The experimental results support our basic thesis, that RTL is able to induce social interactive behavior in humans as the HP's initial exploratory behavior toward the robot changed into relating, recognizing, understanding, and going-along with the robots programmed "intentions". A minority of the HP's failed to understand the "invitation to end", which indicates that the chosen robot action was too subtle. What's more, to a large extent, the 'communication' in the interactions was carried out through the HP's body and kinesthetic sense.

VII. Conclusion

Our work with robotic training partners has resulted in technology and experience that support forceful high-bandwidth kinesthetic HRI with a quality that enables and facilitates natural and intuitive bodily interaction between Human Participants (HP) and Robot Participants (RP).

When designing unobtrusive HRI in our augmented gravity simulations, the HP accept the RP as a natural and unnoticeable part of the interaction with their physical world. This has allowed us to alter the natural conditions of physical training, helping the paralyzed to train in spite of gravity and would be astronauts in spite of the lack of gravity. This ability is similar to existing commercial and experimental computerized training and rehab products and projects, as described in section II

When designing obtrusive HRI — simulating an inverse spring or engaging in sequential interaction, the HP accepts the RP as a noticeable interaction partner or opponent, immediately initiating social interaction as evidenced in gaze behavior, expressive body language, and attempts to overcome or explore the intentions of the RP. This ability have allowed us to kinesthetically facilitate simple instructions and increase motivation and effort during physical training experiments.

Perspective and future work

The ability to provoke strong bodily reactions toward the RP will be explored further. In this way we hope to better understand and exploit our findings to improve motivation, effort and results in strength training for the healthy as well as the disabled.

We expect that studying and documenting interaction elements between HP's and obtrusive RP's, as well as exploring the boundary between unobtrusive and obtrusive behavior, will provide valuable insights that will enable us to make computerized training machines more user friendly, increase the outcome of the training, and increase the motivation of the HP, during training.

We are currently designing a series of experiments, where a small set of basic physical exercises are provided by an unobtrusive, as well as an obtrusive RP, chosen at random for each test person. Exposing a large group of HP's to the test will provide data on the correlation between physical effort and the type of interaction.

We are also designing more intricate sequential interactions, in order to explore how state changes between unobtrusive and obtrusive interaction can be used to facilitate intent from RP to HP, as in instructions on how to perform training.

Initially these experiments will be performed with students as HP's, and as the training programs are refined, we will engage patients as test persons. It will be particularly interesting to explore kinesthetic interaction with stroke patients and other neurological patients with reduced cognitive abilities.

Acknowledgment

The authors would like to thank Allan Lauritsen, Jesper Kiersgaard, Danish Aerospace Company, Odense University Hospital, Region Syddanmark, RoboCluster, Energi Fyn, our students technical staff and volunteers, for all their help and support.

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Evaluation of a Device that Combines Exercise and Entertainment -In the view of Electroencephalogram analysis-

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***Abstract*— The purpose of this study is to develop a device that combines exercise and entertainment. Therefore, we developed an application and a driving device to visualize the energy consumption during exercise. The first device utilizes a smart phone; the second devices utilize the action of jumping rope, handgrip, boxercise, or playing a Kendama. The result of the test is that the driving devices responded correctly to the shaking or sound input of the smart phone, as well as the jumping rope action, handgrip action, or a Kendama action. We investigated the physiological effects of this device using EEG as a measurement system. In the examination of the EEG, it was found that after use, a sense of relaxation was obtained.**

Keywords—Exercise and a driving device; Training; Rehabilitation; EEG

I. INTRODUCTION

Physical fitness tests conducted in Japan have been a decreasing trend since the 1980s. As a result, an increasing requirement of medical expenses and a large number of elderly people who need to be taken care of have become serious problems in Japan society [1]. As a cause of the decline in physical strength and exercise ability, 1) exercise time to play outside and sports due to popularization of television and mobile games is getting less, 2) The places for children to play are decreasing, 3) Attending a cram school aimed at higher education deprives much of the children's playing time, and exercise time is decreasing. Therefore, it is important to increase the children's playing time and continuity for physical exercises.

On the other hand, rehabilitation devices have been developed for motor function improvement of the elderly and the handicapped. However, the activities of these instruments which are aimed at certain exercising abilities or muscular recovery tend to be monotonous. In addition, innovation in sensor technology improvement and dissemination of smart phones and tablets of the digital device of the embedded technology by the development of IT technology is progressing rapidly. The smartphone has an acceleration sensor, a temperature sensor, a gravity sensor, and so on. We have maintain and improve physical fitness and health by managing the number of steps using acceleration sensor and managing mileage, time and Consumption calorie with the use of acceleration sensor, GPS, to manage the distance traveled, time, consumption calorie, and to maintain and improve physical strength and health With these devices, it is difficult to simultaneously solve the shortage of exercise, which is a problem among generations, and research and development are being carried out as individual measures at the present.

Therefore, in order to solve these problems, 1) Simple exercises can be performed in a narrow space. 2) Constant motion can be done regardless of age. 3) Continued exercise can be done. 4) Inexpensive application of existing embedded technology is possible.

It is necessary to develop an exercise device which has the above characteristics.

II. MANUFACTURE OF A DEVICE THAT COMBINES EXERCISE AND ENTERTAINMENT

With the device, no matter where children, the elderly, the disabled or ordinary people, anyone can do exercises at the same time. Moreover, it is a set (consisting of applications and driving devices) which can even make comparisons. This enhances entertainment by combining various conventional exercises and training with devices with gaming properties, and there is ease of us so that anyone can easily feel free to play.

Furthermore, this device is characterized in that it is not necessary to unify the exercise in order to compare momentum. Currently, we have developed six types of devices. A) Shuffling the smart phone; B) Sound input like applause; C) driven by rope skipping; D) driven by muscular strength training; E) driven by playing Kendama; F) driven by tapping the screen of the mobile phone; G) driven by playing a boxercise. In this device has possible competition with various exercises by using the same driving device.

A. Application and the driving device using acceleration sensor

Figure 1 is the driving device of the smart phone. The device composes two parts of devices. One is the device for reading the movements. The other one is the driving device in response to the movements. This set of device can read by application the figure of the acceleration sensor. As the communication method, 500msec every count's electricity will be supplied. The driving device is equipped with an Arduino and FET driver to control the motor.

B. Application and the driving device using sound input

Figure 2 shows the application and the driving device which uses sound input. Visual display is possible, in case of a low sound, a small circle in bottom right will be seen as showed in figure 2(a); but if the sound is loud, the circle will turn to a larger one, as what figure 2(b) shows. The subjects of this set of device are the elderly and the disabled, etc. The driving device can be operated when there're claps, sound of instruments or any other voices.

C. Driving device that reacts to a jumping rope action

A jumping rope action and the driving device are showed in Figure 3. This set of device is made up of the movements of a jumping rope action and the driving device which will react to the movements.

D. Driving device that reacts to rehabilitation exercises and grip strength training

Figure 4 is a set of device with a handgrip used for grip strength training or rehabilitation exercises and the driving device. For the method of detecting the "hold" action, at first the maximum or minimum resisting figure of the bending sensor will be detected, then it will be counted when it is within 10% of either of the two figures.

E. Driving device that reacts to Kendama

Figure 5 is the set that contains a color sensor and the driving device. The sensor will be fixed in the side cups (large and small) and the handle cup of the Kendama, times of the ball hitting the large side cup and the handle cup by turns will be counted, thus the motor in the device set can be driven.

F. Driving device that reacts to tap the screen of the mobile phone

Figure 6 is a device in which a drive device moves by tapping a marker placed on a smartphone.

Since this device can adjust the display speed, it can be used from elderly people to children.

G. Driving device that reacts to play a boxercise

Figure 7 is a set of punching gloves and mitts. This device incorporates a pressure sensitive sensor in the mitt and gives the driving device 500 msec per times as it is one pressed.

Figure 8 show the relationship between various exercise input devices and exercise, rehabilitation, and entertainment. As shown in the figure, this device does not only exercise, but also incorporates elements to enjoy.

III. Confirmation and Verification on this Device Movements

In order to observe the use situation of this device, we asked children to use for trial use. Figure 9 is the photograph of the children using this device at the Arakawa Industry Exhibition event. As a result of using it in the event, about 30

Figure 1 Device driven by application using acceleration sensor and shuffling movement

Figure 2 Device driven by sound input

Figure 3 Device driven by a jumping rope action instrument

Figure 4 Device driven by handgrip instrument

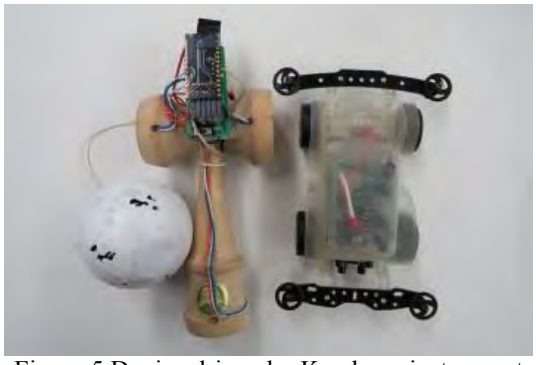

Figure 5 Device driven by Kendama instrument

Figure 6 Device driven by tap the screen of the smartphone

Figure 7 Device driven by hit a mitt

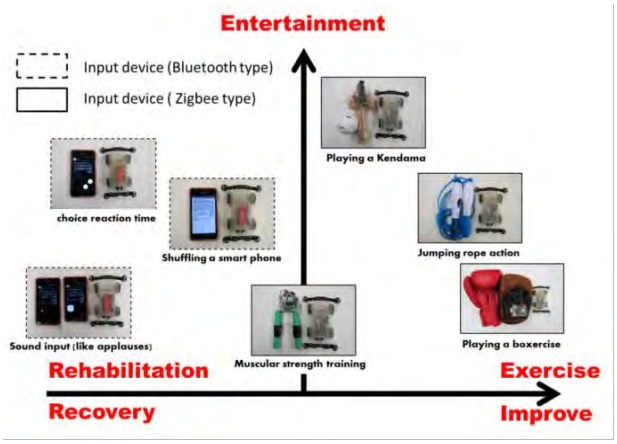

Figure 8 Relationship of devices and communication type

Figure 9 Image of arakawa industry exhibition event

Figure 10 Image of the home for the aged event

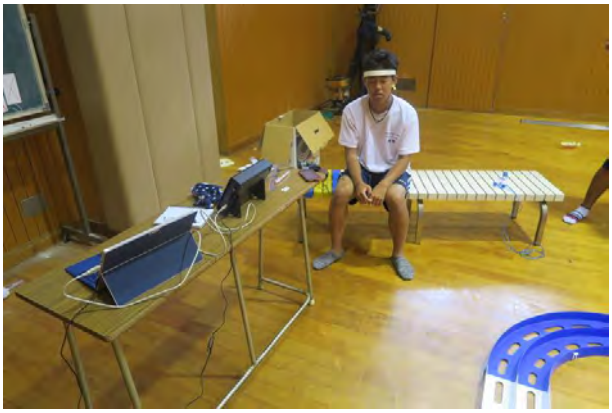

Figure 11 Experiment of EEG

people were able to exercise with enjoyment using this device. Furthermore, in order to ascertain whether this device can be used even by elderly people, I used it in a home for the aged. There were a total of 20 subjects, among them included 10 mild dementia patients and 10 severe dementia patients.

Figure 10 shows the situation when a patient with mild dementia uses this device. In the case of patients with mild dementia, they were able to understand and exercise on how to play this device. However, in the case of patients with severe dementia, they were unable to understand this device. This device was found to be capable of corresponding to the mild authentication patient.

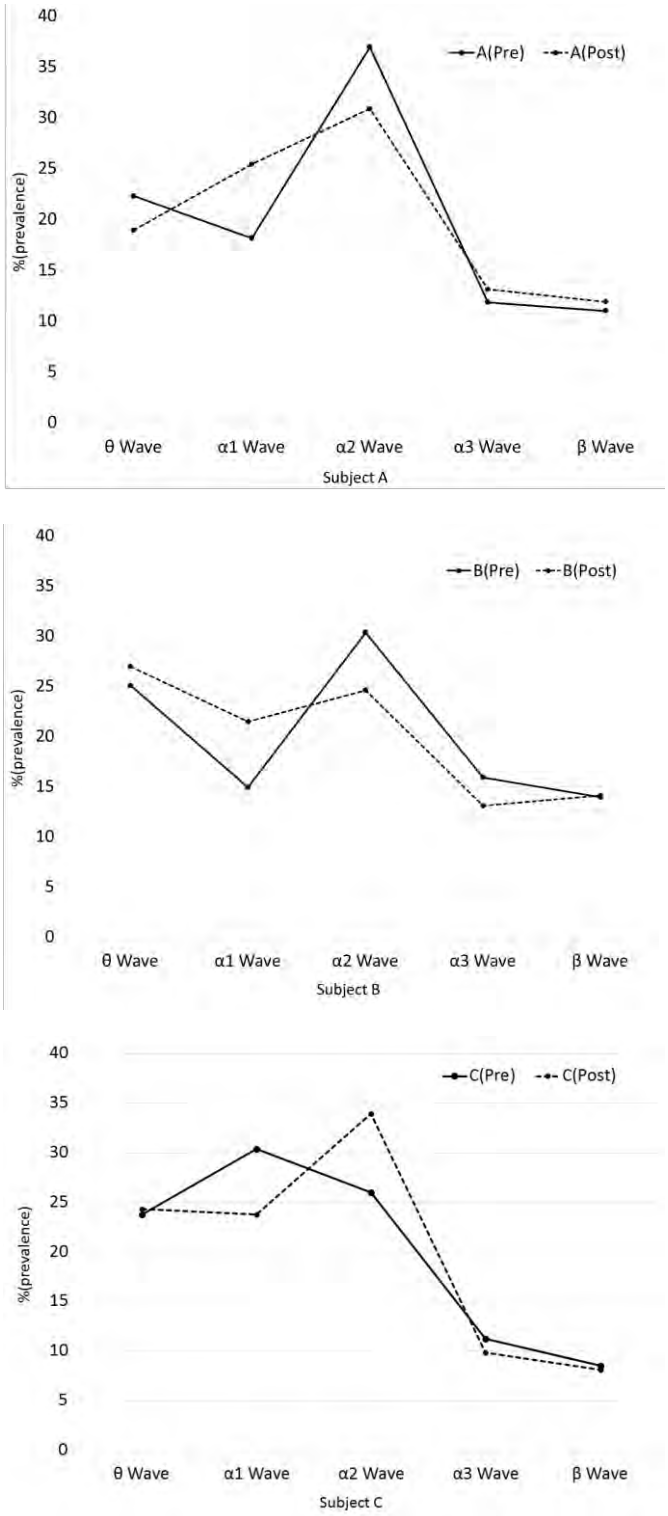

Figure 12 Results of The EEG measurement

IV. RESULT AND METHOD OF MEASURING EEG

To investigate physiological effects with this device, we used electroencephalography (EEG) as measurement system tools. The subjects were five male students (15-17 years old). Figure 11 is the measurement landscape of the measurement experiment of brain waves. Measurement directed a subject to sit down on a chair so as not to move a head.

The EEG (FM-717; Futek Electronics Co. Ltd.) was recorded using a monopolar lead from Fp1 and Fp2 according, with an Ag/AgCl electrode and referenced to the left earlobe (A1). Artefacts such as ocular movement were removed from the EEG data. The power density of each of θ (5.0Hz; sleepiness state), α1 (7.5Hz; relax state), α2 (10.0Hz; relax and concentrate state), α3 (12.5Hz; concentrate and tense state), and β (22.0Hz; dispersion state of concentration because of the tense state) band taken in condition was averaged and used for analysis. The EEG (FM-717; Futek Electronics Co. Ltd.) was recorded using a monopolar lead between Fp1 and Fp2 according, with an Ag/AgCl electrode and referenced to the left earlobe (A1). EEG was recorded for one minute each before and after exercise. That exercise was a move to shuffle smartphones for the measurement of EEG. The course of experiment was one lap for 4 meters, drive device ran four laps.

Figure 12 shows the results of three subjects. Subject A used α 2 waves and α 1 waves decreased after using the device. Subject B had a high relax and concentrate state, but sleepiness state and relax state increased after the exercise. This may be because exercise intensity was too low. Therefore, it is thought that there was little exercise stimulation and drowsiness became high. Subject C shows a tendency to relax and to have a concentration state after the exercise though the relaxation state was high. As a result of the experiment, it was suggested that the relaxation effect and concentration effect could be obtained by using this device.

V. CONCLUSION

In this study, we developed a drive unit to convey the device and the motion to detect the body movement.

By developing motion input that combines mobile phones or exercise device, it became a highly entertaining device. For children and elderly people, it was possible to enjoy using the device.

Moreover, in the examination of the EEG, it was found that after use, a sense of relaxation was obtained. This device can to use from children for exercise to the elderly people for rehabilitation.

For future plans, we are to increase the number of subjects and examine to detail.

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Stride length estimation in self-pace and brisk walking with a single inertial measurement unit on the shank

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***Abstract*— The use of inertial sensors to measure spatial parameters of human gait is becoming more and more attractive because of their low cost and simple use in settings outside the laboratories and long tracking experiments. Furthermore, several wearable exoskeletons embed and use such sensors in the internal control loop or to measure the user training progress by monitoring the spatiotemporal parameters of gait, such as the stride length.**

Many algorithms for stride length estimation have been developed. However, it is fundamental to implement methodologies to identify critical points on stride length estimation, depending on different walking patterns. In this paper, we studied the effect on stride length estimation of three different techniques to remove the gravitational component of acceleration in self-pace and brisk walking by using an inertial measurement unit on the lateral side of shank above the ankle. We evaluate the techniques by considering precision, accuracy, and shape of the histogram of the stride estimation error.

I. Introduction

In the last decade, several wearable exoskeletons for paraplegic patients have been developed, such as HAL [1] and ReWalk [2]. However, the success of walking support systems is based on how easily the user can interface with the wearable robot. The interface should be able to automatically determine important walking parameters such as starting and stopping of walking, stride length, direction, and so on [3].

Inertial sensors are a great advantage for wearable robots because they are small, compact, lightweight, and low energy. Therefore, they are embedded in most of the current exoskeletons [4], [5].

Kagawa and Al. [6] proposed a control scheme for their wearable robot in which the length estimation of next stride is based on the measurement of the current stride length using inertial sensors and foot pressure sensors. Brescianini and Al. [7] also measure stride length using inertial sensors embedded in the crutches. Furthermore, wearable exoskeletons have been used in post-stroke rehabilitation of gait and some researchers found that sub-acute stroke patients might experience added benefit from exoskeletal gait training compared to standard rehabilitation [8].

For a proper stride control during training protocols the spatiotemporal gait characteristics must be monitored and stride length is surely one of the most important [9]–[11]. Wearable exoskeletons are used for ground walking and, differently than treadmill-based robots, it is more difficult to measure the stride length based only on the kinematic of the exoskeleton.

The use of inertial sensors to measure spatial and temporal parameters of human gait is becoming more and more attractive because of the low cost of the technology and its simple use in non-controlled settings outside the laboratories and for long continuous tracking [12]. Besides, several studies demonstrated that inertial sensors are reliable tools for the calculation of temporal gait parameters in different conditions of walking [13]–[15]. However, the estimation of spatial parameters, such as stride length, is a more complex problem. This problem has been addressed in different ways using kinematic and dynamic models [16] or behavioral considerations on the foot motion such as in the Zero Velocity Update (ZUPT) algorithm [17]. However, the above-mentioned algorithms for stride length estimation rely on 1) precise gait events detection; 2) estimation of the net acceleration of body segments after removing gravitational components. While the first point has been widely explored because it is strongly related with the reliability of temporal parameters, more work is required to clarify the effect of different techniques to remove gravitational components in the estimation of stride length. Many algorithms for stride length estimation already exist in literature [18]–[20], and it is fundamental to develop methodologies to identify critical points on stride length estimation depending on different walking patterns, and find solutions to minimize estimation errors. To address the above problem, this paper aims to evaluate the results of different techniques of stride length estimation by using inertial measurement unit on the shank, for different walking patterns. In particular, we perform the evaluation in the case of self-pace and brisk walking for three different techniques used to remove the gravitational component of the acceleration.

II. Material and Method

A. Inertial Measurement Unit

Fig. 1.a shows the 9-axis inertial measurement unit named Waseda Bioinstrumentation Rev. 4 (WB-4) that includes an

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accelerometer, a gyroscope, and a magnetometer [21]. A Bluetooth interface allows wireless data transmission for a maximum distance of 10 meters at frequency rate of 400 Hz. Sensors are secured on the subject's shanks with an adjustable elastic band close to the ankle to minimize artifacts (Fig. 1.b).

B. Motion capture system

The experiments have been performed in a room equipped with a Raptor-E motion capture system (sample frequency 60Hz) with two reflective markers placed on the right and left side of the subject along the line that connects the opposite sides of the anklebone (Fig. 1.c). We evaluated the stride length measuring the distance between the position of the marker at the stance phase and the next position at the stance phase. We excluded in the calculation the first and the last stride to remove effects due to acceleration/deceleration phases. The measurements of the motion capture system have been regarded as ground-truth in the evaluation of the performance of the algorithms for stride length estimation using IMUs.

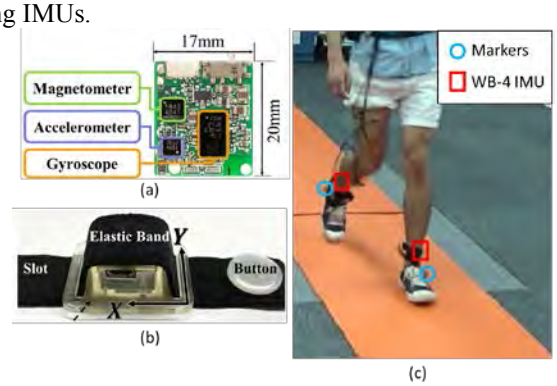

Fig. 1: a) WB-4 Inertial Measurement Unit (IMU); b) Band to secure WB-4 on the shank; c) Position of WB-4 and reflective markers.

C. Experimental Protocol

In this study, we asked seven male healthy subjects (age: 22-24 years old; weight: 61.2±9.2 kg; height: 1.70±0.05 m; BMI: 21.2±2.98) to walk straight a distance of 5 m at usual self-pace for five times. After, we asked to repeat the procedure walking at brisk pace. The subjects wore their normal shoes during the experiment.

D. Stride length estimation

The algorithm for stride length estimation consists of the following steps:

1) *Swing phase estimation:* the angular velocity is used to determine the initial and terminal contact of the foot according to the algorithm described in [15].

2) *Net acceleration calculation:* the sensor attitude is calculated from the sensor data using the following methods.

Method A) The attitude during the stance phase uses accelerometer data. The attitude during the swing phase is calculated by integration of the angular velocity measured by the gyroscope.

Method B) The Extended Kalman Filter (EKF) proposed in [22] estimates the attitude of the sensor in the global reference system.

Method C) R-adaptive EKF proposed in [23] has been used for the attitude estimation of the sensor in a global reference system. The block diagram of the algorithm is shown in Fig 2

After the sensor attitude is estimated, the net acceleration is obtained by using the following equation:

$$\vec{a} = {}^{IMU}_{0}R\vec{a}_{IMU} - \begin{bmatrix} 0 \\ 0 \\ g \end{bmatrix} \qquad (1)$$

Where ${}^{IMU}_{0}R$ is the rotation matrix from the sensor (IMU) to the global (0) reference system; $\vec{a}_{IMU}$ is the acceleration data on the sensor reference system; $\vec{a}$ is the net acceleration in the global reference system. The North-East-Down (NED) has been used as global reference system.

4) *ZUPT algorithm:* we used the original method described in [24] to obtain the displacement vector $p = [p_x \quad p_y \quad p_z]$ during the swing phase in the global reference system.

5) *Stride length calculation:* we calculated the stride length considering only the components on the plane x-y:

$$stride\ length = \sqrt{p_x^2 + p_y^2} \qquad (2)$$

III. Results

We calculated the histograms of stride estimation error for the three different methods for both self-pace and brisk walking with bin size 0.1 m in the range [-1, 1] m.

The *stride estimation error* is the difference between the stride length measured in the motion capture system and the correspondent value estimated by the algorithm. Table I shows the characteristics of the strides.

TABLE I: STRIDES ANALYZED AND CHARACTERISTICS

Walking type	# of strides	Stride length (σ)
Self-pace	180	1.310 (0.213) m
Brisk	156	1.457 (0.375) m

Fig. 2 shows the histogram of the error of the estimated stride length for Method A. The histograms have the highest peak for both self-pace and brisk walking around -0.1 m but the precision of the estimation is poor (self-pace: σ =0.379 m; brisk: σ =0.507 m). Furthermore, especially in the case of brisk walking (Fig. 2.b), the histogram shows secondary peaks and in general the stride length is under-estimated (average error: -0.319m). The histogram of the error for Method B shows a better precision (self-pace: σ =0.130; brisk: σ =0.182) but lack of estimation accuracy (Fig. 3). The algorithm systematically under-estimates the stride length of about 0.45 m in average (self-pace: 0.443 m; brisk: 0.475 m). Finally, Fig. 4 shows the effect of Method C on the estimation error of the stride length. Despite the precision is slightly worse than Method B (self-pace: σ =0.196m; brisk: σ =0.243) the whole estimation accuracy improved with a residual bias of about -0.07 m (self-pace: -0.052 m; brisk: -0.089 m).

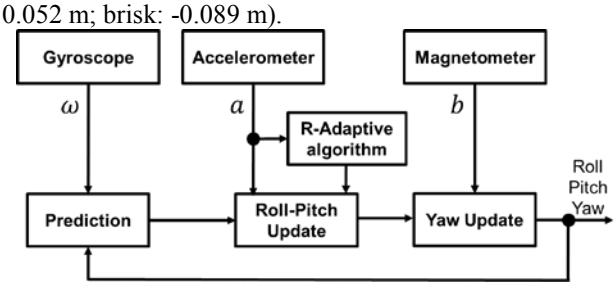

Fig. 2: block diagram of the R-adaptive EKF. $\boldsymbol{\omega}$ is the angular velocity, $\boldsymbol{a}$ is the acceleration, and $\boldsymbol{b}$ is the magnetic field measured by the sensor.

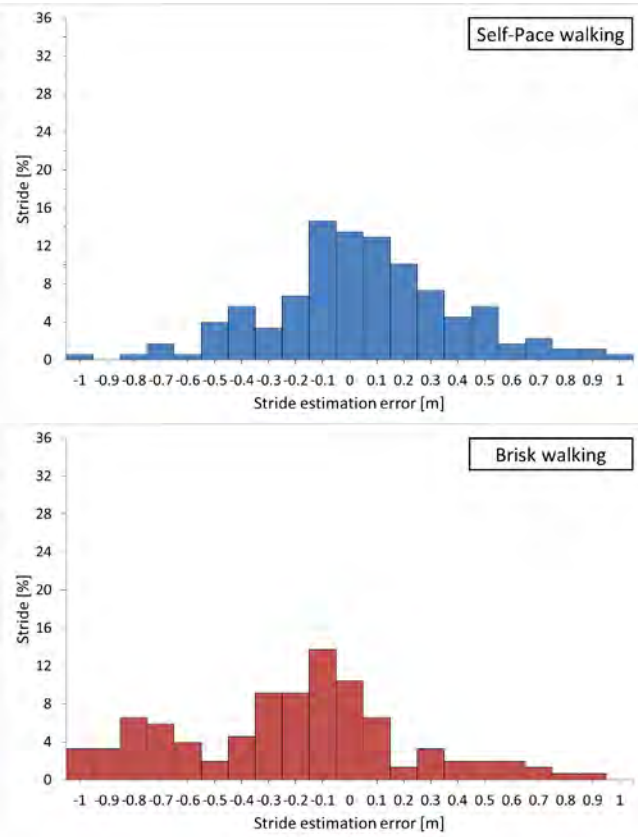

Fig 3: Histogram of the errors for Method A. Self-pace (blue) brisk (red)

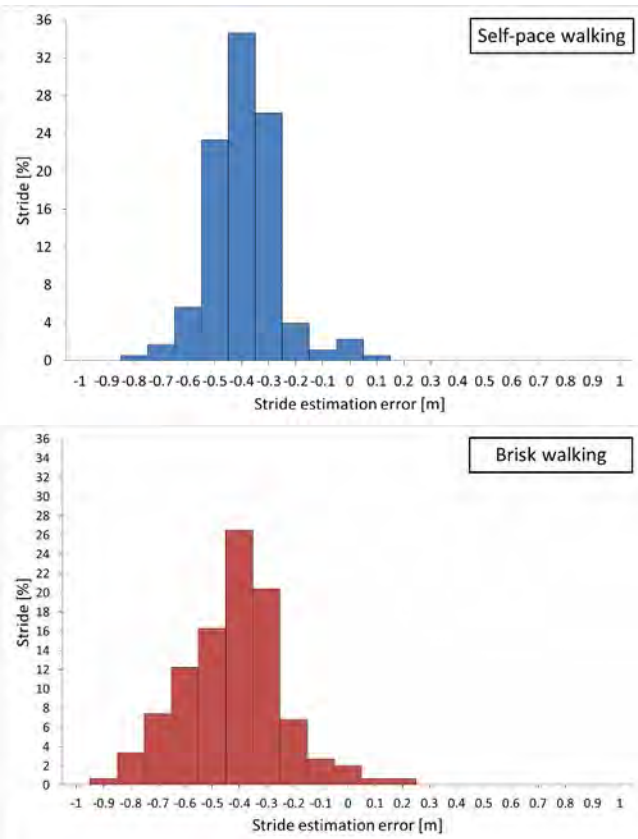

Fig 4: Histogram of the errors for Method B. Self-pace (blue) brisk (red)

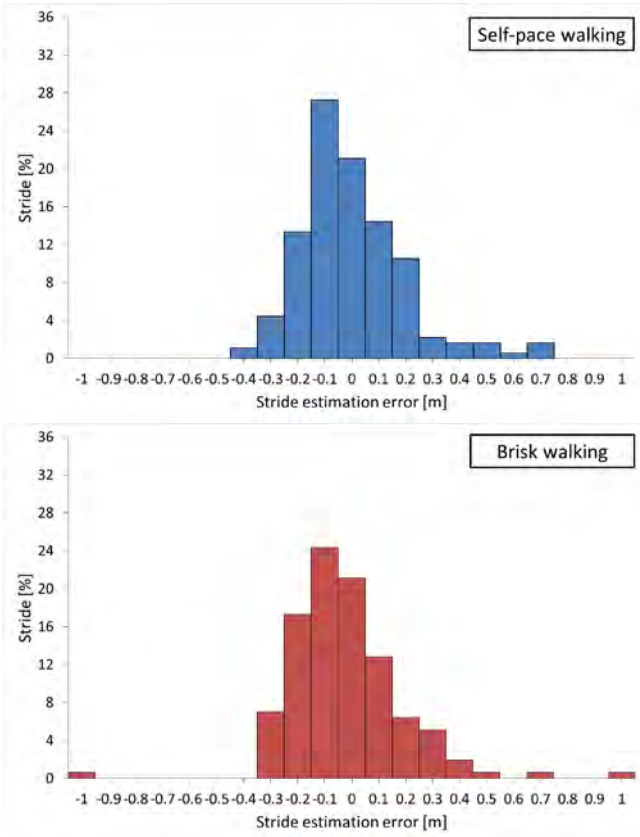

Fig 5: Histogram of the errors for Method C. Self-pace (blue) brisk (red)

IV. Conclusions

In this paper, we studied the effect on stride length estimation of three different techniques to remove the gravitational component of acceleration in self-pace and brisk walking by using an inertial measurement unit on the shank. Among the three methods, we observed that the use of an R-adaptive Extended Kalman Filter (EKF) for the attitude estimation of the sensor provides a smaller negative bias estimation and a near-Gaussian distribution of the stride estimation error. A possible explanation of the negative bias is that the integration in the ZUPT is performed only during the swing phase but during the stance phase the shank still moves due to the rotation of the ankle joint. Further experiments placing the sensor on the foot are needed to verify this hypothesis. Our results suggest that estimation of stride length is better during normal walking than during brisk walking regardless of the algorithm used to obtain the net acceleration. Therefore, other reasons such as sample rate and event detection should be considered to understand this particular behavior. Furthermore, our results show that methods to remove gravitational components are critical for precise estimation of stride length and must be selected carefully. Although this is a pilot study, it is very useful to verify the feasibility and to identify the criticality on the stride length estimation from IMU data. In future works we are planning to make an extensive experiment campaign with more subjects and to consider different methods for event estimation.

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An Interaction Analysis of User-Testing to Extract Salient User Experience with the Robotic Assistive Device *Life-Walker*

Takumi Ohashi[1], Makiko Watanabe[2], and Miki Saijo[1]

***Abstract*— A methodology for real environment user-testing aiming at extracting salient user experience is demonstrated in this paper. Attribution theory, which is a cognitive psychological theory, is applied to analyze the interaction of users.**

I. Introduction

The proportion and absolute number of elderly people in populations around the world are increasing dramatically. In particular, Japan is the only country where the proportion of people aged 60 years or older exceeded 30% in 2015. However, by the middle of the century, many countries will have a proportion of elderly people similar to Japan's in 2015 [1]. Thus, robotic assistive devices, which would reduce individual and societal costs and enable ageing in place, are attracting growing attention [1, 2]. It is known that physical activity lowers the risks of cognitive impairment, Alzheimer disease, and dementia of any type [3], but the obstacles for frail elderly to engage in physical activity are very high.

We have, so far, studied how frail elderly people can go on enjoyable outings with the help of a robotic assistive device called Life-Walker (LW), a 4-wheel electric assisted bicycle (see Fig. 1 and our previous research [4]). The concept of this machine is not only to encourage physical activity, but also to enhance quality of life (QoL). It is generally agreed that social interactions play a beneficial role in maintaining psychological well-being [5], and we are hopeful that the device will function to mediate communication with people around the user. Needless to say, just presenting elderly users with the assistive device is not enough. We need to find ways to prod them to go on outings and otherwise help them to mitigate negative experiences.

Because user experience is a consequence of a user's internal state, the characteristics of the designed system and the context within which the interaction occurs [6], the user test for extracting user experience should be conducted in the context of the real-life environment in which the device is intended to be used.

We conducted a one-day real-environment workshop in Kakegawa, Shizuoka Prefecture, Japan, in which both healthy and frail elderly users enjoyed an outing together using the LW, and then we analyzed the interaction between them. To analyze this interaction, we employed attribution theory, which is a cognitive psychological theory. We conducted a questionnaire survey and retrospective interviews as well to achieve greater objectivity. In this report, we would like to introduce the methodology of the user-testing as well as parts of the analysis results to demonstrate how we extracted salient user experiences through interaction analysis and retrospective interviews. At the conference, we would like to demonstrate the quantitative analysis of the interactions and extraction of salient user experiences, and discuss the factors for prodding frail elderly people to go on outings.

II. Attribution Theory

In the areas of motivation and emotion in achievement, Weiner's attribution theory of achievement motivation and emotion has long attracted attention. In achievement-related contexts, Weiner proposed three dominant causal perceptions: *locus of causality* (causes perceived as residing with or outside of the person), *stability* (causes perceived as stable or unstable), and *controllability* (causes that the agent may or may not change). He claimed that all three dimensions of causality affect a variety of common emotional experiences, including anger, gratitude, guilt, hopelessness, pity, pride, and shame. Therefore, depending on the ascription of the events or outcomes, a person's expectancy and affect will change, resulting in a directing of the motivated behavior [7, 8]. As Weiner himself noted, *controllability* is distinguishable from *locus of causality* and *stability* yet also overlaps with or may not be empirically orthogonal to these other two dimensions [8]. In this report, therefore, we omitted this dimension and employed the *locus of causality* and *stability* dimensions in a 2×2 representation [9] including four determinants of behavioral outcomes (*ability*, *effort*, *task difficulty*, and *luck*), as shown in Table I.

TABLE I. Classification Scheme for the Perceived Determinants of Achievement Behavior [9]

Locus of causality	**Stability**	
	Stable	***Unstable***
Internal	Ability	Effort
External	Task difficulty	Luck

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III. METHODOLOGY

User-testing and data analysis were carried out as explained below:

Date: Nov. 19, 2014.

Location: 22 Century Hill Park, Kakegawa, Shizuoka, Japan

Subjects: Six pairings of frail elderly persons and healthy elderly persons

Because of space limitations, in this report we discuss only one of the pairings, frail elderly person F1 and healthy elderly person H1.

Methods: Questionnaire survey, Voice recording of the pairs' interaction while test-riding the Life-Walker, and retrospective interviews of each pair.

Timeline:

Health check (15 min)

Guidance (15 min)

1st LW test ride (10–15 min)

1st retrospective interview on the ride (5 min)

2nd LW test ride (10–15 min)

2nd retrospective interview on the ride (5 min)

Questionnaire survey (5 min)

Analysis:

Amount of data to be analyzed

Conversation in 1st test ride: 2490 turns* (97 min) / 6 pairs

1st retrospective interviews: 18 min

Conversation in 2nd test ride: 1638 turns (76 min) / 6 pairs

2nd retrospective interviews: 38 min

*One turn lasts until the partner speaks

Procedures

1. Pre-treatment of recorded voices
   i. Divide the recorded voices by turn
2. Ascribe each user utterance into the categories of *luck*, *task difficulty*, *ability*, or *effort*
3. Elucidate the salient events for users by observing the shift of attribution
4. Evaluate the salient user experience using the results of attribution shift and retrospective interviews

The utterances were translated from Japanese to English by a professional translator and the authors.

Figure 1. Test riding of LW; F1 and H1 pair.

IV. RESULTS AND DISCUSSIONS

A. Utterance analysis using attribution theory

Here, we present a partial outline of the utterance analysis of the 1st test ride by the F1-H1 subject pair. The number shown is the turn number, the brackets indicate the speaker. The text below is an English translation of the original transcription in Japanese.

Excerpt 1-1
153. [F1] Just as I thought. Can't do it.
155. [F1] When I first practiced on this, you know, my left leg, you know,
157. [F1] didn't have an artificial bone yet.

This utterance is of impossibility due to an external constraint (i.e. an artificial bone). This excerpt can be ascribed to *task difficulty*.

Excerpt 1-2
167. [F1] Yeah, yeah. You know, I want to skim along, [get the wheels going] swish, swish, swish.
168. [H1] Right. You want to get this thing moving, I know.
173. [F1] Yeah, right. See, see how high I can get my leg up.
176. [H1] Yeah, you are getting it up. Oh, but just one foot [on the pedal] isn't going to be enough, is it.
177. [F1] Could I do it with just one foot? You think?
178. [H1] Don't know about that.
179. [F1] Well, well, I could try.
180. [H1] Give it a try. That's it, that's right.
181. [F1] I can't get it back though.
182. [H1] Yeah, you can't get it back, can you.

H1's utterances seem to be accepting of F1's utterances. Encouraged by this acceptance, F1 tries to control the pedal with only one foot but is unable to do so. F1 appears to recognize that the obstacle is her own lack of ability. This excerpt can be ascribed to *ability*.

Excerpt 1-3
278. [H1] Well then, you've gotten the hang of this.
279. [F1] Yes, I have.

From the point of H1's utterance at turn no. 278, F1's utterances dramatically change and self-monitoring utterances of her control over the driving start to increase (see Excerpt 1-4).

Excerpt 1-4
357. [F1] Go this way.
359. [F1] No, not just a little bit.
361. [F1] Change direction.
363. [F1] That's it. Now let go again.
365. [F1] Let go just a little and change direction.

Self-monitoring of driving control indicates that she recognizes the results are up to her (*locus of causality*: internal) and success or failure will depend on her paying close attention (*stability*: unstable), therefore, this excerpt can be ascribed to *effort*.

Excerpt 1-3 also indicates that F1 is strongly affected by H1's utterances; she is, in other words, affected by the interaction with H1, leading us to the conclusion that the salient user experience for F1 is her interaction with H1. We will discuss this further in the next section from the perspective of the retrospective interviews.

B. Retrospective interview

After the LW test ride, we conducted retrospective interviews of the participants. A couple of the questions and responses in the 2nd retrospective interview are shown here:

Q. How was it walking together? What did you like about it?

[F1] The person assisting me kept telling me how good I was (laughter). I got so carried away I even honked at the person in front of me (laughter).

Q. Did it make you happy being helped in this way?

[F1] You know how you end of repeating the same things over and over. When the person assisting you is really good at it, you feel like you got a good grade, you know.

This shows how the interaction with H1 was important to F1. In addition, one of the retrospective interview questions was, "Did participating in this event change your feelings in any way?" Give your response on a 5-point scale (5: Changed – 1: Not at all）. F1 selected 5 (Changed), saying, "It was fun to drive and enjoy the scenery. At home, I might be able to walk only about 300 meters using a cane. I enjoy being able to pedal on my own." Clearly, F1's impressions were positively changed.

It should be noted here that F1 was not so positive in the 1st retrospective interview, a fact that needs further study.

C. Implications to be drawn from a partial analysis

Fig. 2 shows F1's attribution shift from *task difficulty* to *effort*, resulting primarily from her interaction with H1. Also, the 2nd retrospective interview indicates a shift to positive impression and improved motivation. This phenomenon can be explained by prior studies that report a positive correlation between achievement motivation and the ascribing of success to a high degree of *effort* [10].

Locus of causality	**Stability**	
	Stable	***Unstable***
Internal	Ability	Effort
External	Task difficulty	Luck

F1: (Positively) Changed

Figure 2. Reattribution and interview results.

V. Conclusion

In this paper, we have introduced the methodology of our user testing and presented a partial analysis result. It may be possible to extract salient user experience by coding utterances based on attribution theory and observing shifts in attribution.

At the conference, we will report on our analysis of multiple samples and demonstrate the quantitative analysis of the participants' utterances. Also, we will discuss the merits and limitations of this methodology for user testing, as well as how frail elderly persons can be prodded to go on enjoyable outings.

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