NEW RESEARCH FOR MEMS IN THE VESTIBULAR SYSTEM AND BASIC OVERVIEW OF MEMS USED TILL NOW

V. Manoev, Ts. Kachamachkov

Technical University of Sofia, Bulgaria
Faculty of Telecommunication, TU-Sofia, "Kl. Ohridsky" str. 8, 1000 Sofia, Bulgaria

Abstract

The ear is the organ used for hearing and maintaining equilibrium. In the inner ear, the vestibular system is responsible for the sense of balance. The main organs of the vestibular system are the semicircular canals, the saccule, and the utricle. Within each of the vestibular organs, sensory receptors in the form of hair cells detect motion and send a message to the brain for interpretation. This paper presents a feasibility study on an MEMS – based implantable vestibular solutions. Such as the MEMS - Implantable vestibular prosthesis that will replace the function of the damaged vestibular end-organ by providing an MEMS chip that will replace the function of the damaged vestibular end-organ by providing an MEMS chip that will accurately sense, extract, and transmit 3-dimensional motion information for people who have permanently lost peripheral vestibular function. The prosthesis prototype includes orthogonal triads of accelerometers and gyroscopes on a single fingernail sized chip. Based on the physiological data on the perceptual thresholds of linear acceleration and angular velocity in humans.

- *New research for mems in the vestibular system.*
- *The author has made a thorough research of existing researches.*
- *Achieving a thorough overview of current researches for mems in the vestibular system.*

1. INTRODUCTION

This paper investigates an ear worn sensor for the development of a gait analysis framework. Instead of explicitly defining gait features that indicate injury or impairment, an automatic method of feature extraction and selection is proposed. The proposed framework uses multi-resolution wave let analysis and margin based feature selection. It was validated on three datasets; the first simulating a leg injury, the second simulating abdominal impairment that could result from surgery or injury and the third is a dataset collected from a patient during recovery from leg injury. The method shows a clear distinction of gait between injured and normal walking. It also illustrates the fact that using source separation before pattern classification can significantly improve the proposed gait analysis framework.

2. USIND HEAD WORN ACCELERATION SENSORS

Features that can quantify human gait play a major role in the studies of injury rehabilitation, improving athletic performance, and the design of prosthetic limbs. Human gait has long been an active area of research, and many systems have been proposed for observing gaiter regularities. Thus far, most of these systems are based on image information from video sequences. When used in conjunction with biomechanical models, these features can allow quantitative analysis of many specific gait characteristics such as joint moments and powers (kinetic analysis), joint angles, angular velocities, and angular accelerations(kinematic analysis) [1]. In these systems, optical markers are placed near anatomical landmarks of the body and features related to gait are extracted from video sequences. Parametric models have been used extensively to describe a set of image observations. An example is 3-D modelling of moving people which can be achieved by using volumetric bodies based on elliptical cylinders [2] or tapered super quadrics. Alternatively, 2-D models that represent the projection of 3-D data to an imaging plane can be used (2D/3D contour-modelling for example [3]).In general, methods that consider the use of kinematic constraints can handle more complex motions and occlusions. These constraints are used to improve tracking as well as preventing model violations of the derived 3-D structures. For example, Dockstader et al. [4] use soft constraints in a hierarchical structural model of the human body to analyse video sequences captured in a home environment. Reviews on human motion analysis based on video, including summaries of modelling, tracking and recognition, are provided by Dariush [5] and Aggrawal et al.[6].An alternative to vision based gait analysis is the use of body worn sensors to obtain motion data. The variables that can be measured during gait analysis depend on the technique selected. The most commonly measured variables include initial contact (IC) that defines the beginning of a complete gait cycle and thus cycle duration and frequency, and terminal contact (TC) that marks the start of the swing phase. Gyroscopes, which measure angular velocity, and accelerometers which measure linear acceleration, have been used as a wearable option to measure these variables [7] .Coleman et al. [8], Aminian et al. [7] and Selles et al. [9] provide methods of measuring both TC and IC timing information. On the other hand, Yoshida et al. [10] use an accelerometer attached to the patient's waist and observe frequency peaks in the anterior plane to detect leg injury. In reality, however, injury causes changes in both the temporal and frequency domains and this variation is not normally limited to one direction of motion. Due to leg injury, a patient's motion can show signs of swaying for maintaining equilibrium, as well as a change in the distribution of the forces when the patient's foot touches the floor. Injury indifferent parts of the body, resulting from abdominal surgery for example, could also affect gait and balance, especially when a patient needs to climb stairs, reach for objects or liedown.

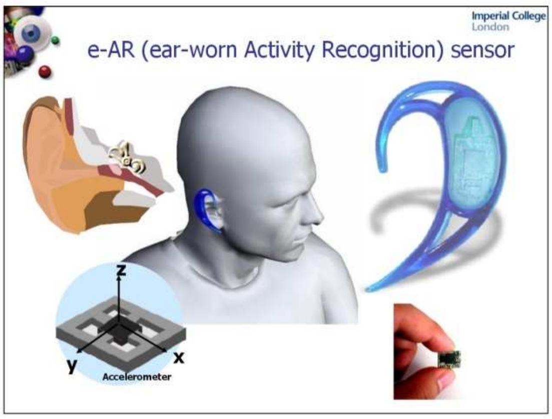

Figure 1: e-AR sensor

3. EXPERIMENTAL SETUP

Head-worn accelerometers have in the past been used to study the movement of the head compared to the rest of the trunk, as well as gait changes due to disease and aging. As a subject mobilises, the head remains relatively stable when compared to the trunk, and therefore the direction of its movement during activities such as walking is more representative of the body's movement as a whole. A study by Kavanagh et al (2005) [11] used tri-axial accelerometers attached to the head and trunk of normal volunteer subjects to collect data whilst they were walking, and showed that accelerations detected at the head were not only more regular than those at the trunk in each direction, but also associated with a greater degree of coupling between directions [12]. Authors concluded therefore that during walking, accelerations of the head are significantly attenuated and more tightly controlled when compared to accelerations of the lower trunk. Despite being a relatively stable part of the body during motion, the head moves proportionately more with increasing activity (for example climbing stairs versus walking on a flat level) [13], which is important to consider when evaluating gait. It does however move less than the limbs due to the stabilising effect conferred by the trunk and neck, making it an attractive site for sensor placement [14,15]. The trunk does play a critical role in main taining head stability by regulating gait-related oscillations in all directions, and the neck confers additional control the rebyenhancing head stability during walking [11].The experiments outlined in this work are designed to observe changes in gait that occur following impaired truncal and limb immobility as detected by an ear-worn activity recognition sensor (e-AR sensor) [16]. The e-AR sensor has been used previously for observing post-operative recovery [16], activities of daily living and sports performance. Amulti-resolution wavelet-based framework is proposed in this paper to analyse the e-AR sensor's accelerometer data in the 3 directions of motion. Automatic margin-based feature selection is used to select frequency bands of the wavelet transform that provide discrimination between classes of motion. Independent Component Analysis (ICA) is used as ameans of source separation for 3-D acceleration.

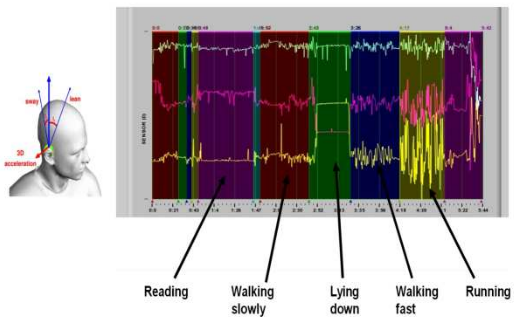

Figure 2: e-AR sensor experimental results.

The purpose of the semicircular canal prosthesis is to restore balance function. Ideally, the prosthesis will be able to sense motion with suficient precision and to deliver signals to the central neural system matching signals that the natural organ would generate.We propose to implement the vestibular prosthesis using MEMS technology. An ensemble of six inertial MEMS sensors is required to measure six-degrees of freedom of the head motion on a single silicon chip and packaged in a volume smaller than 1 cubic centimeter. Micromachining can shrink the sensors size by orders of magnitude, reduce the fabrication cost significantly, and allow the electronics to be integrated on the same siliconchip [17]. Our first prototype of the vestibular prosthesis is implemented using polysilicon surface micromachining technology. In the prosthesis, the three semicircular canals are replaced by 3-axis MEMS gyroscopes, while the two otolith organs are replaced by 3-axisacceleromters, Fig-3.

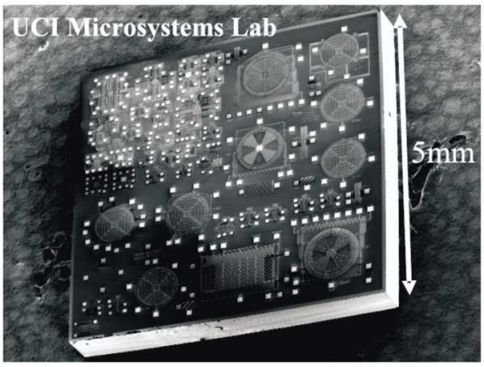

Figure 3. A prototype multi-sensor unit including accelerometers and gyroscopes. The experimental unit does not include electronics on the chip. The experimental chip is fabricate dusing JDS/Uniphase's MUMPs technology

The MEMS accelerometer consists of a proof mass suspended by compliant beams anchored to a xed frame. External acceleration due to motion of the object to which the sensor's frame is attached, displaces the support frame relative to the proof mass, which in turn, changes the initial stress in the suspension spring. Both this relative displacement and the suspension-beam stress scan be used as a measure of the external acceleration. In the most general case, the proof-mass motion can have six degrees of freedom. But typically in a unidirectional accelerometer, the geometrical design of the suspension is such that one of these axes has low stiftness while high stifness along other axes. For example, in case of the Z-axis accelerometer, the proof mass of the device will displace in out-of-plane of the chip only if there is an acceleration component along the z-axis. In the basis of operation, the proof-mass, which constitute the active portion of the sensor, is driven by an oscillator circuit at a precise amplitude and high frequency. When subjected to a rotation, the proof-mass will be subjected to the Coriolis force:

$$F = 2m\Omega\, x\, V_c \qquad (1)$$

where m – mass, Vc – instantaneous radial velocity of the center of mass, – input rate. As an illustrative example, consider a Z-axis gyroscope. The behaviour of the gyroscope is naturally described Schematic illustration of a MEMS implementation of the z-axis rate integrating gyroscope with respect to the non-inertial coordinate frame {x; y; z},

$$X'' + \omega_n^2 X - 2\Omega Y' = 0 \qquad (2)$$

$$Y'' + \omega_n^2 Y - 2\Omega X' = 0 \qquad (3)$$

The essential feature of these equations is the presence of the Coriolis acceleration terms$-2\Omega Y'$ and $-2\Omega X'$It is the Coriolis acceleration that causes a transfer of energy between the two gyroscope modes of operation. The resultant Coriolis force is perpendicular to both the input rate and the instantaneous radial velocity in the drive direction. This produces a motion of the proofmassin direction perpendicular to its initial oscillation. Under rotation, however, the Coriolis acceleration will cause energy to be transferred from the x-axis(primary mode) to they-axis (secondary mode) building up a vibration amplitude along the y-axis. The ratio of the amplitude in the secondary mode of vibration to the amplitude of the primary mode of vibration can be shown to be proportional to the rotation rate and is given by [18]

$$\frac{Y}{X} = 2Q\frac{\Omega}{\omega_n} \qquad (4)$$

CONCLUSION

This work was carried out in conjunction with SAPHE (Smart and Aware Pervasive Healthcare Environment)supported by the UK Technology Strategy Board (TSB) and WASP (Wirelessly Accessible sensor populations), funded by the European Commission under Framework 6. We would also like to thank all subjects who took part in this study.

ACKNOWLEDGMENTS

The research described in this paper is supported by the Scientific Research Sector of TU-Sofia under the contract No 152пд0054-07

References

[1] G.F. Harris and J.J. Wertsch, Procedures for Gait Analysis. Arch Physical Medicine and Biology, 1994. 75(2): p. 216-225.

[2] D. Hogg, Model based vision: A program to see a walking person. Image Vis. Comput., 1983. 1(1): p.5-20.

[3] M. Isard and A. Blake, Condensation-Conditional density propagation for visual tracking, Int. J.Comput. Vis., 1998.29(1): p. 5-28.

[4] S.L. Dockstader, M.J. Berg, and A.M. Tekalp, Stochastic kinematic modeling and feature extraction for gait analysis. Image Processing, IEEE Transactions on, 2003. 12(8): p. 962-976.

[5] B. Dariush, Human motion analysis forbio mechanics and biomedicine. Machine Vision and Applications, 2003. 14(4):p. 202-205.

[6] J. K. Aggrawal, and Q. Cai, Human motion Analysis: Areview. Comput. Vis. Image Understanding, 1999. 73(3): p.428-440.

[7] K. Aminian, et al., Temporal feature estimation during walking using miniature accelerometers: an analysis of gait improvement after hip arthroplasty. Med Biol Eng Comput, 1999. 37(6): p. 686-91.

[8] K. L. Coleman et al., Step Activity Monitor: Long Term-Continuous Recording of Ambulatory Functions. J. Rehab. Res. Develop., 1999. 36(1): p. 8-18.

[9] R. W. Selles et al., Automated estimation of initial and terminal contact timing using accelerometers; development and validation in transtibial amputees and controls. Neural Systems and Rehabilitation Engineering, 2005. 13(1): p. 81-88.

[10] T. Yoshida, et al. Gait Analysis for Detecting a leg accident with an accelerometer. in 1stDistributed Diagnosis and Home Healthcare (D2H2) Conference,. 2006. Virginia, USA.

[11] J. Kavanagh, R.S. Barrett, and S. Morrison, The role of the neck and trunk in facilitating head stability during walking. Exp Brain Res, 2006. 172(4): p. 454-63.

[12] J. Kavanagh, S. Morrison, and R.S. Barrett, Coordination of head and trunk accelerations during walking. Eur J Appl Physiol, 2005. 94(4): p. 468-75.

[13] R. Cromwell and R. Wellmon, Sagittal plane head stabilization during level walking and ambulation on stairs. Physiother Res Int, 2001. 6(3): p. 179-92.

[14] E. Hirasaki et al., Analysis of head and body movements of elderly people during locomotion. Acta Otolaryngol Suppl, 1993. 501: p. 25-30.

[15] H. B. Menz, S. R. Lord, and R. C. Fitzpatrick, Acceleration patterns of the head and pelvis when walking on level and irregular surfaces. Gait Posture, 2003. 18(1): p. 35-46.

[16] B. Lo, L. Atallah, O. Aziz, M. ElHelw, A. Darzi, G-Z. Yang Real-time pervasive monitoring for postoperative care. Proc. of BSN07, IFMBE, vol. 1, 2007; 122–127.

[17] A. Shkel. Smart mems: Micro-structures with error-suppression and self-calibration control capabilities. The American Control Conference, June 2001. Arlington, VA (Invited).

[18] A. Shkel, R. Horowitz, A. Seshia, S. Park and R. T. Howe. Dynamics and control of micromachined gyroscopes. The America Control Conference, June 1999. San Diego, CA.