



US010631740B2

(12) **United States Patent**  
**Romesburg**

(10) **Patent No.:** **US 10,631,740 B2**  
(45) **Date of Patent:** **Apr. 28, 2020**

(54) **REDUCTION OF PHYSIOLOGICAL METRIC ERROR DUE TO INERTIAL CADENCE**

(58) **Field of Classification Search**  
CPC ..... A61B 5/024; A61B 5/721; A61B 5/112;  
A61B 5/0205; A61B 5/021  
See application file for complete search history.

(71) Applicant: **Valencell, Inc.**, Raleigh, NC (US)

(72) Inventor: **Eric Douglas Romesburg**, Chapel Hill, NC (US)

(56) **References Cited**

(73) Assignee: **VALENCELL, INC.**, Raleigh, NC (US)

U.S. PATENT DOCUMENTS

(\* ) Notice: Subject to any disclaimer, the term of this patent is extended or adjusted under 35 U.S.C. 154(b) by 0 days.

3,636,617 A	1/1972	Schmidt et al.
3,704,706 A	12/1972	Herczfeld et al.
4,672,976 A	6/1987	Kroll
4,830,014 A	5/1989	Goodman et al.
4,952,928 A	8/1990	Carroll et al.
4,955,379 A	9/1990	Hall

(Continued)

(21) Appl. No.: **16/694,402**

FOREIGN PATENT DOCUMENTS

(22) Filed: **Nov. 25, 2019**

CN	1545979 A	11/2004
CN	101317188 A	12/2008

(65) **Prior Publication Data**

US 2020/0085314 A1 Mar. 19, 2020

(Continued)

**Related U.S. Application Data**

OTHER PUBLICATIONS

(63) Continuation of application No. 16/146,362, filed on Sep. 28, 2018, now Pat. No. 10,542,896, which is a continuation of application No. 14/370,689, filed as application No. PCT/US2012/071594 on Dec. 24, 2012, now Pat. No. 10,349,844.

Pierpont, G. et al., "Assessing Autonomic Function by Analysis of Heart Rate Recovery from Exercise in Healthy Subjects", The American Journal of Cardiology, Vo. 94, Jul. 1, 2004, pp. 64-68, Excerpta Medica, Inc.

(Continued)

(60) Provisional application No. 61/586,884, filed on Jan. 16, 2012.

*Primary Examiner* — Mohamed Charioui

(74) *Attorney, Agent, or Firm* — Coats & Bennett, PLLC

(51) **Int. Cl.**

<i>A61B 5/024</i>	(2006.01)
<i>A61B 5/00</i>	(2006.01)
<i>A61B 5/11</i>	(2006.01)

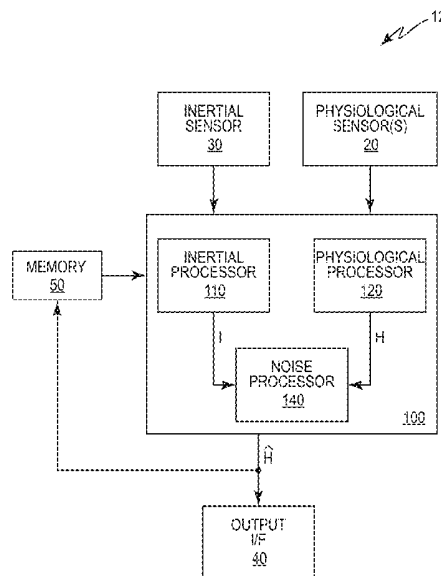
(57) **ABSTRACT**

The heart rate monitor disclosed herein removes a step rate component from a measured heart rate by using one or more filtering techniques when the step rate is close to the heart rate. In general, a difference between the step rate and the heart rate is determined, and the step rate is filtered from the heart rate based on a function of the difference.

(52) **U.S. Cl.**

CPC ..... *A61B 5/024* (2013.01); *A61B 5/112* (2013.01); *A61B 5/6817* (2013.01); *A61B 5/721* (2013.01); *A61B 2562/0219* (2013.01); *A63B 2220/17* (2013.01); *A63B 2230/04* (2013.01)

**18 Claims, 10 Drawing Sheets**



(56)

References Cited

U.S. PATENT DOCUMENTS

5,139,025 A 8/1992 Lewis et al.  
 5,226,417 A 7/1993 Swedlow et al.  
 5,243,992 A 9/1993 Eckerle et al.  
 5,297,548 A 3/1994 Prologe  
 5,299,570 A 4/1994 Hatschek  
 5,396,893 A 3/1995 Oberg et al.  
 5,448,082 A 9/1995 Kim  
 5,482,036 A 1/1996 Diab et al.  
 5,494,043 A 2/1996 O'Sullivan et al.  
 5,503,016 A 3/1996 Koen  
 5,575,284 A 11/1996 Athan et al.  
 5,673,692 A 10/1997 Schulze et al.  
 5,797,841 A 8/1998 Delonzor et al.  
 5,817,008 A 10/1998 Rafert et al.  
 5,807,267 A 11/1998 Bryars et al.  
 5,846,190 A 12/1998 Woehrle  
 5,853,005 A 12/1998 Scanlon  
 5,906,582 A 5/1999 Kondo et al.  
 5,908,396 A 6/1999 Hayakawa et al.  
 5,941,837 A 8/1999 Amano et al.  
 5,954,644 A 9/1999 Dettling et al.  
 5,964,701 A 10/1999 Asada et al.  
 6,022,748 A 2/2000 Charych et al.  
 6,042,549 A 3/2000 Amano et al.  
 6,067,462 A 5/2000 Diab et al.  
 6,198,394 B1 3/2001 Jacobsen et al.  
 6,241,684 B1 6/2001 Amano et al.  
 6,267,721 B1 7/2001 Welles  
 6,393,311 B1 5/2002 Edgar, Jr. et al.  
 6,443,890 B1 9/2002 Schulze et al.  
 6,527,711 B1 3/2003 Stivoric et al.  
 6,608,562 B1 8/2003 Kimura et al.  
 6,656,151 B1 12/2003 Blatter  
 6,702,752 B2 3/2004 Dekker  
 6,721,584 B2 4/2004 Baker, Jr. et al.  
 6,725,072 B2 4/2004 Steuer et al.  
 6,745,061 B1 6/2004 Hicks et al.  
 6,748,254 B2 6/2004 O'Neil et al.  
 6,801,799 B2 10/2004 Mendelson  
 6,898,451 B2 5/2005 Wuori  
 6,995,665 B2 2/2006 Appelt et al.  
 6,996,427 B2 2/2006 Ali et al.  
 6,997,879 B1 2/2006 Turcott  
 7,018,338 B2 3/2006 Vetter et al.  
 7,107,088 B2 9/2006 Aceti  
 7,113,815 B2 9/2006 O'Neil et al.  
 7,144,375 B2 12/2006 Kosuda  
 7,190,986 B1 3/2007 Hannula et al.  
 7,209,775 B2 4/2007 Bae et al.  
 7,336,982 B2 2/2008 Yoo  
 7,378,954 B2 5/2008 Wendt  
 7,438,688 B2 10/2008 Kobayashi et al.  
 7,539,533 B2 5/2009 Tran  
 7,962,308 B2 6/2011 Makino  
 8,055,469 B2 11/2011 Kulach et al.  
 8,109,874 B2\* 2/2012 Kong ..... A61B 5/0205  
 600/300  
 8,923,941 B2 12/2014 LeBoeuf et al.  
 9,005,129 B2 4/2015 Venkatraman et al.  
 9,044,180 B2 6/2015 LeBoeuf et al.  
 9,717,412 B2\* 8/2017 Roham ..... A61B 8/0866  
 2002/0013538 A1 1/2002 Teller  
 2003/0065269 A1 4/2003 Vetter et al.  
 2003/0109791 A1 6/2003 Kondo et al.  
 2003/0176815 A1 9/2003 Baba et al.  
 2003/0181798 A1 9/2003 Al-Ali  
 2003/0233051 A1 12/2003 Verjus et al.  
 2003/0236647 A1 12/2003 Yoon et al.  
 2004/0004547 A1 1/2004 Appelt et al.  
 2004/0039254 A1 2/2004 Stivoric et al.  
 2004/0054291 A1 3/2004 Schulz et al.  
 2004/0059236 A1 3/2004 Margulies et al.  
 2004/0186387 A1 9/2004 Kosuda et al.  
 2004/0186695 A1 9/2004 Aoshima et al.  
 2004/0178913 A1 11/2004 Penuela et al.

2004/0236233 A1 11/2004 Kosuda et al.  
 2004/0242976 A1 12/2004 Abreu  
 2004/0254501 A1 12/2004 Mault  
 2005/0007582 A1 1/2005 Villers et al.  
 2005/0043652 A1 2/2005 Lovett et al.  
 2005/0059870 A1 3/2005 Aceti  
 2005/0075542 A1 4/2005 Goldreich  
 2005/0192516 A1 9/2005 Takiguchi et al.  
 2005/0209516 A1 9/2005 Fraden  
 2005/0212405 A1 9/2005 Negley  
 2005/0228463 A1 10/2005 Mac et al.  
 2005/0245839 A1 11/2005 Stivoric et al.  
 2006/0064037 A1 3/2006 Shalon et al.  
 2006/0084879 A1 3/2006 Nazarian et al.  
 2006/0178588 A1\* 8/2006 Brody ..... A61B 5/02405  
 600/513  
 2006/0258927 A1 11/2006 Edgar, Jr. et al.  
 2007/0016086 A1 1/2007 Inukai et al.  
 2007/0027367 A1 2/2007 Oliver et al.  
 2007/0116314 A1 5/2007 Grilliot et al.  
 2007/0123763 A1 5/2007 Al-Ali et al.  
 2007/0135717 A1 6/2007 Uenishi et al.  
 2007/0197881 A1 8/2007 Wolf et al.  
 2008/0076972 A1 3/2008 Dorogusker et al.  
 2008/0081972 A1 4/2008 Debreczeny  
 2008/0132798 A1 6/2008 Hong et al.  
 2008/0133699 A1 6/2008 Craw et al.  
 2008/0146890 A1 6/2008 LeBoeuf et al.  
 2008/0154098 A1 6/2008 Morris et al.  
 2008/0177162 A1 7/2008 Bae et al.  
 2008/0200774 A1 8/2008 Luo  
 2008/0269625 A1 10/2008 Halperin et al.  
 2009/0010461 A1 1/2009 Klinghult et al.  
 2009/0023556 A1 1/2009 Daly  
 2009/0097689 A1 4/2009 Prest et al.  
 2009/0105556 A1 4/2009 Fricke et al.  
 2009/0112111 A1 4/2009 Shimizu  
 2009/0131761 A1 5/2009 Moroney, III et al.  
 2009/0281435 A1 11/2009 Ahmed et al.  
 2009/0287067 A1 11/2009 Dorogusker et al.  
 2009/0306736 A1 12/2009 Dobak, III  
 2010/0189209 A1 7/2010 O'Rourke  
 2010/0274109 A1\* 10/2010 Hu ..... A61B 5/02405  
 600/310  
 2011/0022352 A1 1/2011 Fujita et al.  
 2011/0178759 A1 7/2011 Uchida  
 2012/0197093 A1 8/2012 LeBoeuf et al.  
 2012/0303319 A1 11/2012 Kirkeby  
 2013/0006583 A1 1/2013 Weast et al.

FOREIGN PATENT DOCUMENTS

CN 101910846 A 12/2010  
 CN 101980659 A 2/2011  
 CN 102168986 A 8/2011  
 CN 102297701 A 12/2011  
 CN 102435203 A 5/2012  
 EP 0729726 A2 9/1996  
 EP 1354553 A1\* 10/2003 ..... A61B 5/02416  
 EP 1354553 A1 10/2003  
 EP 2229880 A1 9/2010  
 EP 2182839 B1 10/2011  
 JP H110258039 A 9/1998  
 JP 2004283228 A 10/2004  
 JP 2004358271 A 12/2004  
 JP 2005040261 A 2/2005  
 JP 2005270544 A 10/2005  
 WO 0021435 A1 4/2000  
 WO 0044274 8/2000  
 WO 2005010568 A2 2/2005  
 WO 2005036212 A2 4/2005  
 WO 2006009830 A2 1/2006  
 WO 2007013054 A1 2/2007  
 WO 2007122375 A2 11/2007  
 WO 2011026669 A 3/2011  
 WO 2011105914 A1 9/2011  
 WO 2013038296 A1 3/2013

(56) **References Cited**

## FOREIGN PATENT DOCUMENTS

WO	2013109389 A1	7/2013
WO	2014109982 A2	7/2014

## OTHER PUBLICATIONS

- Coote, J., "Recovery of Heart Rate Following Intense Dynamic Exercise", *Experimental Physiology—Review Article*, *Exp Physiol* 95.3, Mar. 1, 2010, pp. 431-440, The Physiological Society.
- Wise, K., "Integrated sensors, MEMS, and microsystems: Reflections on a fantastic voyage," *Sensors and Actuators A*, vol. 136, Feb. 5, 2007, pp. 39-50.
- Gigoi, B.P., et al., "Integration Technology for MEMS Automotive Sensors," 28th Annual Conference of the IEEE, Jan. 1, 2002, pp. 2712-2717.
- Ko, W., "Trends and frontiers of MEMS," *Sensors and Actuators A*, vol. 136, Feb. 1, 2007, pp. 62-67.
- Barbour, N., "Inertial Sensor Technology Trends," *IEEE Sensors Journal*, vol. 1, No. 4, Dec. 1, 2001, pp. 332-339.
- Vigario, R., "Independent Component Approach to the Analysis of EEG and MEG Recordings," *IEEE Transactions on Biomedical Engineering*, vol. 47, No. 5, May 1, 2000, pp. 589-593.
- Mayer-Kress, G., "Localized Measures for Nonstationary Time-Series of Physiological Data," *Integr. Physiol. Behav. Sci.*, vol. 29, No. 3, Jul. 1, 1994, pp. 205-210.
- Shaw, G.A., et al., "Warfighter Physiological and Environmental Monitoring: A Study for the U.S. Army Research Institute in Environmental Medicine and the Soldier Systems Center," *Lincoln Laboratory, Massachusetts Institute of Technology, Lexington, MA.*, Nov. 1, 2004, pp. 1-128.
- Laguna, P., et al., "Power Spectral Density of Unevenly Sampled Data by Least-Square Analysis: Performance and Application to Heart Rate Signals," *IEEE Transactions on Biomedical Engineering*, vol. 45, No. 6, Jun. 1, 1998, pp. 698-715.
- Richardson, J.E., "Physiological Responses of Firefighters Wearing Level 3 Chemical Protective Suits While Working in Controlled Hot Environments," *J. Occup. Environ. Med.*, vol. 43, No. 12, Dec. 1, 2001, pp. 1064-1072.
- Scanlon, M., "Acoustic Sensors in the Helmet Detect Voice and Physiology," *Proceedings of SPIE*, vol. 5071, Jan. 1, 2003, pp. 41-51.
- Arnold, M., et al., "Adaptive AR Modeling of Nonstationary Time Series by Means of Kalman Filtering," *IEEE Transactions on Biomedical Engineering*, vol. 45, No. 5, May 1, 1998, pp. 553-562.
- Yan, Y., et al., "Reduction of motion artifact in pulse oximetry by smoothed pseudo Wigner-Ville distribution," *Journal of NeuroEngineering and Rehabilitation*, vol. 2, No. 3, Mar. 1, 2005, pp. 1-9.
- Lee, C.M., et al., "Reduction of Motion Artifacts from Photoplethysmographic Recordings Using a Wavelet Denoising Approach," *IEEE EMBS Asian-Pacific Conference on Biomedical Engineering*, Jan. 1, 2003, pp. 194-195.
- Foo, J.Y.A., "Comparison of wavelet transformation and adaptive filtering in restoring artefact-induced time-related measurement," *Biomedical Signal Processing and Control* vol. 1, No. 1 (2006), Mar. 24, 2006, pp. 93-98.
- Wood, L., et al., "Active Motion Artifact Reduction for Wearable Sensors Using Laguerre Expansion and Signal Separation," *Proceedings of the 2005 IEEE Engineering in Medicine and Biology 27th Annual Conference*, Shanghai, China, Sep. 1, 2005, pp. 3571-3574.
- Cacioppo, J., "Inferring Psychological Significance From Physiological Signals," *American Psychologist*, vol. 45, No. 1, American Psychological Association, Jan. 1, 1990, pp. 16-28.
- Rhee, S., et al., "Artifact-Resistant Power-Efficient Design of Finger-Ring Plethysmographic Sensors," *IEEE Transactions on Biomedical Engineering*, vol. 48, No. 7, Jul. 1, 2001, pp. 795-805.
- Wagner, J., et al., "From Physiological Signals to Emotions: Implementing and Comparing Selected Methods for Feature Extraction and Classification," *IEEE Int. Conf. Multimedia and Expo*, Jan. 1, 2005, pp. 1-4.
- Parkka, J., et al., "Activity Classification Using Realistic Data From Wearable Sensors," *IEEE Transactions on Information Technology in Biomedicine*, vol. 10, No. 1, Jan. 1, 2006, pp. 119-128.
- Georgoulas, G., et al., "Predicting the Risk of Metabolic Acidosis for Newborns Based on Fetal Heart Rate Signal Classification Using Support Vector Machines," *IEEE Transactions on Biomedical Engineering*, vol. 53, No. 5, May 1, 2006, pp. 875-884.
- Liao, W., et al., "A Real-Time Human Stress Monitoring System Using Dynamic Bayesian Network," *Proceedings of the 2005 IEEE Computer Society Conference on Computer Vision and Pattern Recognition*, Jan. 1, 2005, pp. 1-8.
- Moy, M., et al., "Accuracy of uniaxial accelerometer in chronic obstructive pulmonary disease," *Journal of Rehabilitation Research and Development*, vol. 45, No. 4, Nov. 4, 2008, pp. 611-618.
- Moy, M., et al., "Ambulatory Monitoring of Cumulative Free-Living Activity," *IEEE Engineering in Medicine and Biology Magazine* May/Jun. 2003, May 1, 2003, pp. 89-95.
- Ricke, AD, et al. "Automatic Segmentation of Heart Sound Signals Using Hidden Markov Models," *IEEE Computers in Cardiology 2005*; vol. 32, Jan. 1, 2005, pp. 953-956.
- Acharya, R., et al., "Classification of cardiac abnormalities using heart rate signals," *Medical and Biological Engineering and Computing 2004*, vol. 42, No. 3, Jan. 1, 2004, pp. 288-293.
- Allen, F., et al., "Classification of a known sequence of motions and postures from accelerometry data using adapted Gaussian mixture models," *Institute of Physics Publishing Physiological Measurement*, vol. 27, No. 10, Jul. 25, 2006, pp. 935-951.
- Lee, J., et al., "Design of filter to reject motion artifact of pulse oximetry," *Computer Standards & Interfaces*, vol. 26 (2004), Jul. 4, 2003, pp. 241-249.
- Rezek, I.A, et al., "Stochastic Complexity Measures for Physiological Signal Analysis," *IEEE Transactions on Biomedical Engineering*, vol. 45, No. 9, Sep. 1, 1998, pp. 1186-1191.
- Gibbs, P., et al., "Active motion artifact cancellation for wearable health monitoring sensors using collocated MEMS accelerometers," *Smart Struct. Mater.*, International Society for Optics and Photonics, Jan. 1, 2005, pp. 1-9.
- Merletti, R., et al., "Advances in processing of surface myoelectric signals: Part 1," *Medical and Biological Engineering and Computing*, vol. 33, No. 3, May 1, 1995, pp. 362-372.
- Asada, H., et al., "Active Noise Cancellation Using MEMS Accelerometers for Motion-Tolerant Wearable Bio-Sensors," *Proceedings of the 26th Annual International Conference of the IEEE EMBS*, San Francisco, CA, USA, Sep. 1, 2004, pp. 2157-2160.
- Newman, A., et al., "Sleep Disturbance, Psychosocial Correlates, and Cardiovascular Disease in 5201 Older Adults: The Cardiovascular Health Study," *Journal of American Geriatric Society*, vol. 45, No. 1, Jan. 1, 1997, pp. 1-7.
- Chan, K.W., et al., "Adaptive Reduction of Motion Artifact from Photoplethysmographic Recordings using a Variable Step-Size LMS Filter," *IEEE, Sensors*, Jun. 1, 2002, pp. 1342-1346.
- Dew, M.A., et al., "Healthy Older Adults' Sleep Predicts All-Cause Mortality at 4 to 19 Years of Follow-Up," *Psychosomatic Medicine*, vol. 65, Jan. 1, 2003, pp. 63-73.
- Gibbs, P., et al., "Reducing Motion Artifact in Wearable Bio-Sensors Using MEMS Accelerometers for Active Noise Cancellation," *IEEE American Control Conference*, Jun. 1, 2005, pp. 1581-1586.
- Yang, B-H, et al., "Development of the ring sensor for healthcare automation," *Robotics and Autonomous Systems*, vol. 30, Jan. 1, 2000, pp. 273-281.
- Healey, J., et al., "Detecting Stress During Real-World Driving Tasks Using Physiological Sensors," *IEEE Transactions on Intelligent Transportation Systems*, vol. 6, No. 2, Jun. 1, 2005, pp. 156-166.
- Hayes, M.J., et al., "A New Method for Pulse Oximetry Possessing Inherent Insensitivity to Artifact," *IEEE Transactions on Biomedical Engineering*, vol. 48, No. 4, Apr. 1, 2001, pp. 452-461.
- Wilson, G., et al., "An Analysis of Mental Workload in Pilots During Flight Using Multiple Psychophysiological Measures," *The International Journal of Aviation Psychology*, vol. 12, No. 1, May 1, 2001, pp. 3-18.
- Baddeley, A.D., "Selective Attention and Performance in Dangerous Environments," *HPEE*, vol. 5, No. 1, Oct. 1, 2000, pp. 86-91.

(56)

## References Cited

## OTHER PUBLICATIONS

- Wilson, G.F., et al., "Performance Enhancement with Real-time Physiologically Controlled Adaptive Aiding," Proceedings of the IEA 2000 / HFES 2000 Congress, vol. 44, Jul. 30, 2000, pp. 61-64.
- Skinner, M.J., et al., "Workload Issues in Military Tactical Airlift," The International Journal of Aviation Psychology, vol. 12, No. 1, May 1, 2001, pp. 79-93.
- Helander, M., "Applicability of Drivers' Electrodermal Response to the Design of the Traffic Environment," Journal of Applied Psychology, vol. 63, No. 4, Jan. 18, 1978, pp. 481-488.
- Perala, C.H., "Galvanic Skin Response as a Measure of Soldier Stress," Army Research Laboratory, ARL-TR-4114, May 1, 2007, pp. 1-35.
- Zhai, J., et al., "Stress Detection in Computer Users Based on Digital Signal Processing of Noninvasive Physiological Variables," Conf Proc IEEE Eng Med Biol Soc., New York, NY, Aug. 31, 2006; pp. 1355-1358.
- Zhai, J., et al., "Stress Recognition Using Non-invasive Technology," FLAIRS Conference, Melbourne Beach, Florida, May 11, 2006, AAAI Press, pp. 395-401.
- Endler, J., "Signals, Signal Conditions, and the Direction of Evolution," The American Naturalist, vol. 139, Supplement, Mar. 1, 1992, pp. S125-S153.
- Sadeh, A., "The role of actigraphy in sleep medicine," Sleep Medicine Reviews, vol. 6, No. 2, Jan. 1, 2002, pp. 113-124.
- Mendelson, Y. et al., "Skin Reflectance Pulse Oximetry: In Vivo Measurements from the Forearm and Calf," Journal of Clinical Monitoring, vol. 7, No. 1, Jan. 1, 1991; pp. 7-12.
- Konig, V. et al., "Reflectance Pulse Oximetry—Principles and Obstetric Application in the Zurich System," Journal of Clinical Monitoring and Computing, vol. 14, No. 6, Aug. 1, 1998; pp. 403-412.
- Tremper, K. et al., "Pulse Oximetry," Medical Intelligence Article, Anesthesiology, vol. 70, No. 1, Jan. 1, 1989; pp. 98-108.
- Haahr, R. et al. "A Wearable 'Electronic Patch' for Wireless Continuous Monitoring of Chronically Diseased Patients," Proceedings of the 5th International Workshop on Wearable and Implantable Body Sensor Networks, Jun. 1-3, 2008; pp. 66-70.
- Asada, H. et al., "The Ring Sensor: A New Ambulatory Wearable Sensor for Twenty-Four Hour Patient Monitoring," Proceedings of the 20th Annual Conference of the IEEE Engineering in Medicine and Biology Society, vol. 20, No. 4, Jan. 1, 1998; pp. 1906-1909.
- Comtois, G. et al., "A Comparative Evaluation of Adaptive Noise Cancellation Algorithms for Minimizing Motion Artifacts in a Forehead-Mounted Wearable Pulse Oximeter," Proceedings of the 29th Annual International Conference of the IEEE EMBS, Lyon, France, Aug. 23, 2007, pp. 1-4.
- Gupta, G. Sen et al., "Design of a Low-cost Physiological Parameter Measurement and Monitoring Device," IMTC 2007—Instrumentation and Measurement Technology Conference, Warsaw, Poland, May 1, 2007, pp. 1-6.
- Lee, R.G. et al. "A Mobile Care System With Alert Mechanism" IEEE Transactions on Information Technology in Biomedicine, vol. 11, Issue 5, Sep. 1, 2007.
- Declaration of Brian W. Anthony, PH.D. in Support of Petition for Inter Partes Review of U.S. Pat. No. 9,044,180, Exhibit 1003, Petition for Inter Partes Review of U.S. Pat. No. 9,044,180; *Apple, Inc. (Petitioner) v. Valencell, Inc. (Patent Owner)*, IPR 2017-01947, filed Aug. 2-17, 2015, pp. 1-153.
- Mendelson, J., et al., "Measurement Site and Photodetector Size Considerations in Optimizing Power Consumption of a Wearable Reflectance Pulse Oximeter," Proceedings of the 25th Annual International Conference of the IEEE EMBS, Engineering in Medicine and Biology Society, Cancun, Mexico, Sep. 17, 2003, pp. 1-4.
- Palder, et al., "Vascular Access for Hemodialysis, Patency rates and Results of Revision," Annals of Surgery, vol. 202, No. 2, Aug. 1, 1985, pp. 1-5.
- Spigulis, J., et al., "Wearable wireless photoplethysmography sensors," Biophotonics: Photonic Solutions for Better Health Care, Proceedings of SPIE, vol. 6991, May 1, 2008, pp. 1-7.
- Sandberg, M., et al., "Non-invasive monitoring of muscle blood perfusion by photoplethysmography: evaluation of a new application," Acta Physiol Scand., vol. 183, No. 4, Dec. 7, 2004, pp. 335-343.
- Sum, K.W., et al. "Vital Sign Monitoring for Elderly at Home: Development of a Compound Sensor for Pulse Rate and Motion," Studies in Health Technology and Informatics, Personalised Health Management Systems, IOS Press, Jan. 1, 2005, pp. 43-50.
- Mendelson, Y., et al., "A Wearable Reflectance Pulse Oximeter for Remote Physiological Monitoring," Proceedings of the 28th IEEE EMBS Annual International Conference New York City, USA, Aug. 30, 2006, pp. 912-915.
- Jung, W., "Chapter H: OP Amp History," Op Amp Applications Handbook, published by Newnes/Elsevier, Jan. 1, 2005, ISBN-0-7506-7844-5, pp. H.1-H.72.
- Texas Instruments, "General Purpose Operational Amplifiers", SLOOSS094B, Nov. 1, 1970, pp. 1-19.
- Schmitt, O., "A simple differential amplifier," Review of Scientific Instruments vol. 8, No. 126, Apr. 1, 1937, American Institute of Physics, pp. 1-3, available at: <http://dx.doi.org/10.1063/1.1752256>.
- Gray, p. et al., "Recent Advances in Monolithic Operational Amplifier Design," IEEE Transactions on Circuits and Systems, vol. CAS-21, No. 3, May 1, 1974, pp. 317-327.
- Horowitz, P., et al., "The Art of Electronics," Second Edition, Cambridge University Press, Jan. 1, 1989, pp. 98-102.
- Petition for Inter Partes Review of U.S. Pat. No. 9,044,180; *Apple, Inc. (Petitioner) v. Valencell, Inc. (Patent Owner)*, IPR 2017-01947, filed Aug. 15, 2017, pp. 1-86.
- Buchanan, T., et al., "Neuromusculoskeletal Modeling: Estimation of Muscle Forces and Joint Moments and Movements From Measurements of Neural Command," J Appl Biomech, vol. 20, No. 4, Nov. 1, 2004, pp. 1-34.
- Stolwijk, J., "Mathematical Models of Thermal Regulation," Annals of the New York Academy of Sciences, vol. 335, No. 1, Jan. 1, 1980, pp. 98-106.
- Wiggs, L. et al., "Sleep patterns and sleep disorders in children with autistic spectrum disorders: insights using parent report and actigraphy," Developmental Medicine and Child Neurology 2004, vol. 46, No. 6, Jan. 1, 2004, pp. 372-380.
- Hastings, P.C., "Symptom burden of sleep-disordered breathing in mild-to-moderate congestive heart failure patients," European Respiratory Journal, vol. 27, No. 4, Jan. 1, 2006, pp. 748-755.
- Carskadon, M., et al., "Chapter 2—Normal Human Sleep: an Overview," Monitoring and staging human sleep, from Principles and practice of sleep medicine, 5th edition, St. Louis: Elsevier Saunders, Jan. 1, 2011, pp. 1-21.
- Critchley, H., "Electrodermal Responses: What Happens in the Brain," The Neuroscientist, vol. 8, No. 2, Jan. 1, 2002, pp. 132-142.
- Lang, P., et al., "Looking at pictures: Affective, facial, visceral, and behavioral reactions," Psychophysiology, vol. 30, No. 3, Apr. 22, 1992, pp. 261-273.
- Soleymani, M., et al., "Affective Ranking of Movie Scenes Using Physiological Signals and Content Analysis," Proc. 2nd ACM Work. Multimed. Semant., Jan. 1, 2008, pp. 1-8.
- Appelhans, B., et al., "Heart Rate Variability as an Index of Regulated Emotional Responding," Review of General Psychology, vol. 10, No. 3, Sep. 15, 2005, pp. 229-240.
- Postma, D.S., et al., "The natural history of chronic obstructive pulmonary disease," European Respiratory Monograph, vol. 38, Jan. 1, 2006, pp. 71-83.
- Bidargaddi, N., et al., "Ambulatory monitor derived clinical measures for continuous assessment of cardiac rehabilitation patients in a community care model," Pervasive Computing Technologies for Healthcare, 2008 Second International Conference on Pervasive Computing Technologies for Healthcare, Jan. 30, 2008, pp. 1-5.
- Hertzman, A., "The Blood Supply of Various Areas as Estimated by the Photoelectric Plethysmograph," Am J. Physiol, vol. 124, Issue 2, Jul. 18, 1938, pp. 328-340.
- Hayes, M., et al., "Artifact reduction in photoplethysmography," Applied Optics, vol. 37, No. 31, Nov. 1, 1998, pp. 7437-7446.
- Page, E., et al., "Physiological approach to monitor patients in congestive heart failure: application of a new implantable device-

(56)

## References Cited

## OTHER PUBLICATIONS

based system to monitor daily life activity and ventilation,” Eurospace, vol. 9, May 3, 2007, pp. 687-693.

Moy, M., et al., “Free-living physical activity in COPD: Assessment with accelerometer and activity checklist,” Journal of Rehabilitation Research & Development, vol. 46, No. 2, Nov. 2, 2009, pp. 277-286.

Bennett, T., et al., “Development of Implantable Devices for Continuous Ambulatory Monitoring of Central Hemodynamic Values in Heart Failure Patients,” Pacing Clin Electrophysiol. Jun. 2005; vol. 28, No. 6, Jun. 1, 2005, pp. 573-584.

Allen, J., “Photoplethysmography and its application in clinical physiological measurement,” Physiological Measurement, vol. 28, Feb. 20, 2007, pp. 1-39.

Webster, J.G. (ed.), “Design of Pulse Oximeters,” Institute of Physics Publishing, Philadelphia, PA, Jan. 1, 1997, pp. 1-134.

Webster, J.G. (ed.), “Design of Pulse Oximeters,” Institute of Physics Publishing, Philadelphia, PA, Jan. 1, 1997, pp. 135-267.

Shevchenko, Y, et al., “90th Anniversary of the Development by Nikolai S. Korotkoff of the Auscultatory Method of Measuring Blood Pressure,” Circulation, vol. 94, No. 2, Jul. 15, 1996, pp. 116-118.

Han, H., et al., “Development of a wearable monitoring device with motion artifact reduced algorithm,” International Conference on Control, Automation and Systems 2007, Oct. 17, 2007, Seoul, Korea, pp. 1581-1584.

Petition for Inter Partes Review of U.S. Pat. No. 8,157,730; *Apple, Inc.* (Petitioner) v. *Valencell, Inc.* (Patent Owner), IPR 2017-01701, filed Jun. 30, 2017, pp. 1-89.

Petition for Inter Partes Review of U.S. Pat. No. 8,652,040; *Apple, Inc.* (Petitioner) v. *Valencell, Inc.* (Patent Owner), IPR 2017-01702, filed Jun. 30, 2017, pp. 1-70.

Petition for Inter Partes Review of U.S. Pat. No. 8,652,040; *Apple, Inc.* (Petitioner) v. *Valencell, Inc.* (Patent Owner), IPR 2017-01703, filed Jun. 30, 2017, pp. 1-79.

Petition for Inter Partes Review of U.S. Pat. No. 8,888,701; *Apple, Inc.* (Petitioner) v. *Valencell, Inc.* (Patent Owner), IPR 2017-01704, filed Jun. 30, 2017, pp. 1-84.

Declaration of Dr. Majid Sarrafzadeh, Exhibit 1003, Petition for Inter Partes Review of U.S. Pat. No. 8,888,701; *Apple, Inc.* (Petitioner) v. *Valencell, Inc.* (Patent Owner), IPR 2017-01704, filed Jun. 30, 2017, pp. 1-109.

Declaration of Brian W. Anthony, PH.D. In Support of Petition for Inter Partes Review of U.S. Pat. No. 8,157,730, Exhibit 1003, Petition for Inter Partes Review of U.S. Pat. No. 8,157,730; *Apple, Inc.* (Petitioner) v. *Valencell, Inc.* (Patent Owner), IPR 2017-01701, filed Jun. 30, 2017, pp. 1-138.

Declaration of Dr. Majid Sarrafzadeh, Exhibit 1003, Petition for Inter Partes Review of U.S. Pat. No. 8,652,040; *Apple, Inc.* (Petitioner) v. *Valencell, Inc.* (Patent Owner), IPR 2017-01703, filed Jun. 30, 2017, pp. 1-87.

Declaration of Dr. Majid Sarrafzadeh, Exhibit 1003, Petition for Inter Partes Review of U.S. Pat. No. 8,652,040; *Apple, Inc.* (Petitioner) v. *Valencell, Inc.* (Patent Owner), IPR 2017-01702, filed Jun. 30, 2017, pp. 1-92.

Asada, H., et al., “Mobile Monitoring with Wearable Photoplethysmographic Biosensors,” IEEE Engineering in Medicine and Biology Magazine, May/Jun. 2003 Issue, May 1, 2003, pp. 28-40.

Wang, L. et al. “Multichannel Reflective PPG Earpiece Sensor with Passive Motion Cancellation,” IEEE Transactions on Biomedical Circuits and Systems, vol. 1, No. 4, Dec. 1, 2007, pp. 235-241.

Bumgardner, W., “Top 8 Walking Speedometers and Odometers”, retrieved on Jun. 18, 2014, retrieved from internet: <http://walking.about.com/od/measure/tp/speedometer.htm>.

Garmin, “Running Watches Heart Rate Monitor”, Swim Watch, Heart Rate Monitors Reviews, Oct. 12, 2010, retrieved from internet: [http://web.archive.org/web/\\*/http://heartratemonitors-reviews.com/category/swim-watch/](http://web.archive.org/web/*/http://heartratemonitors-reviews.com/category/swim-watch/).

Han et al. “Artifacts in wearable photoplethysmographs during daily life motions and their reduction with least mean square based active noise cancellation method,” Computers in Biology and Medicine 42 (2012), Published Dec. 27, 2011, Elsevier Ltd., pp. 387-393.

Poh, Ming-Zher et al., “Non-contact, automated cardiac pulse measurements using video imaging and blind source separation.” Optics Express, vol. 18, No. 10, May 7, 2010, pp. 1-13.

\* cited by examiner

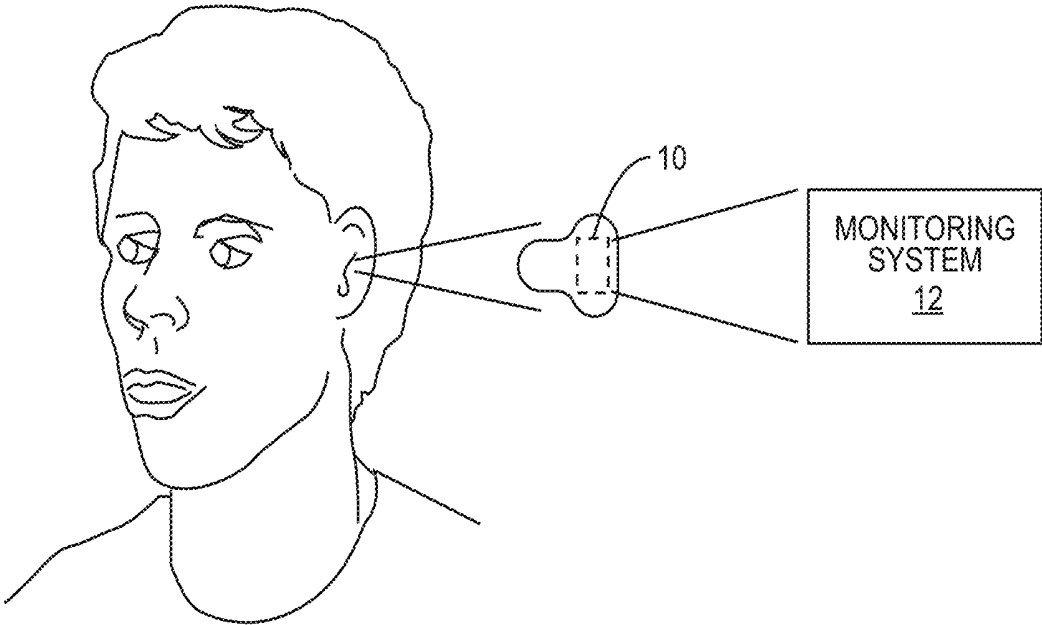


FIG. 1

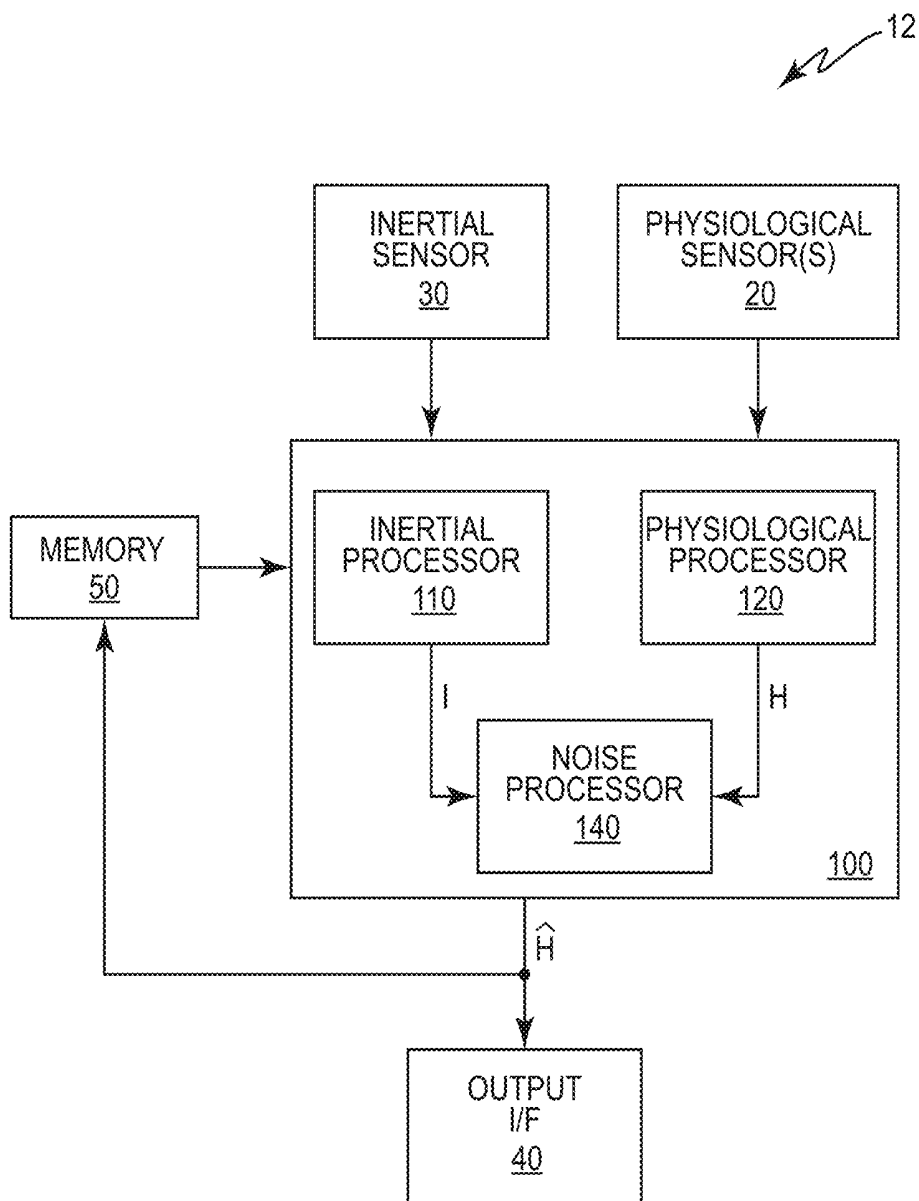


FIG. 2

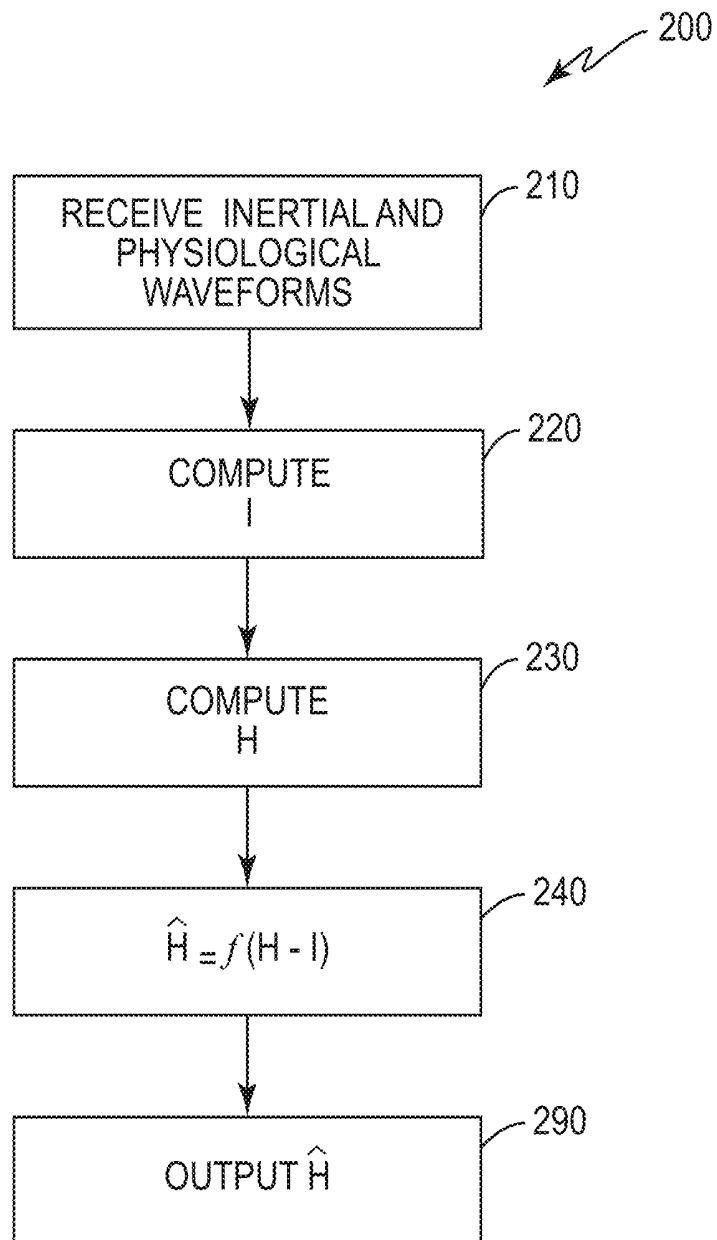


FIG. 3

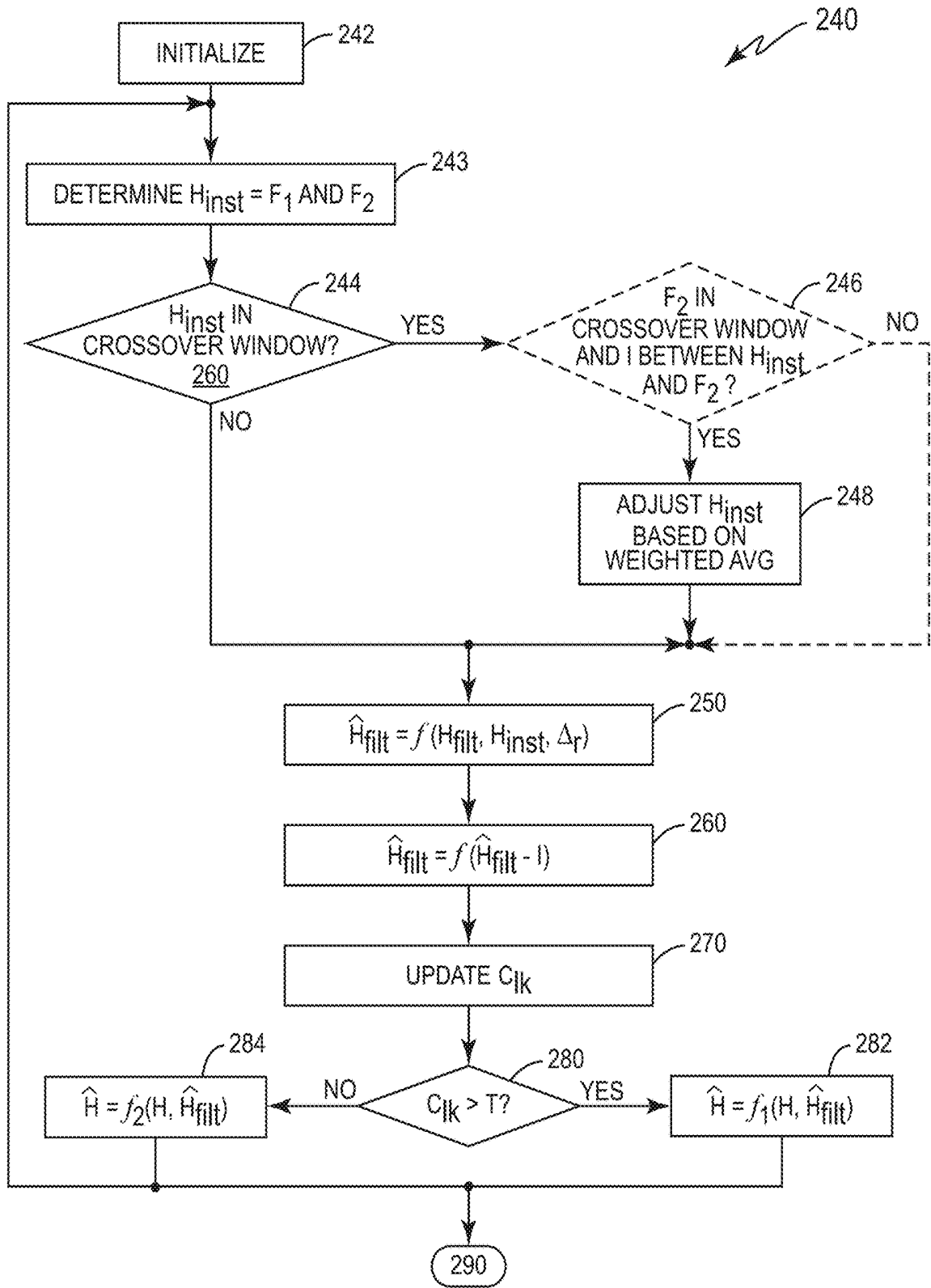


FIG. 4

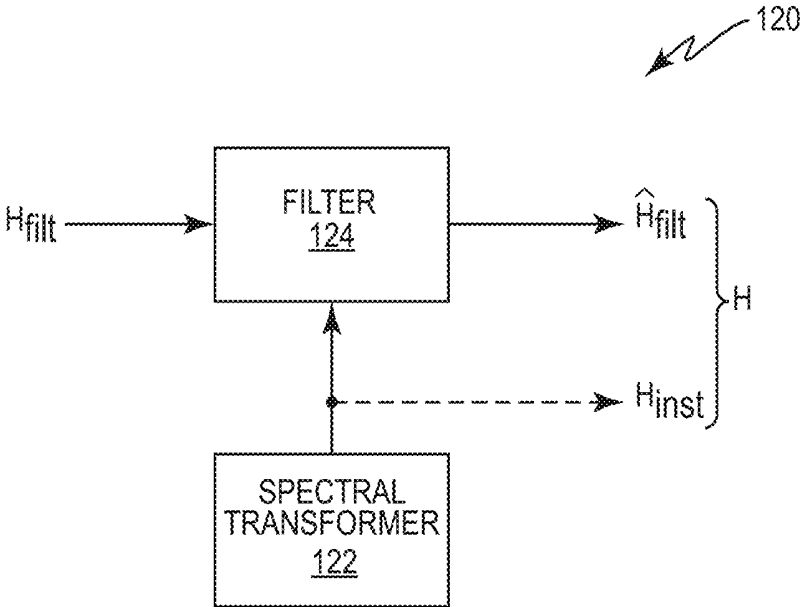
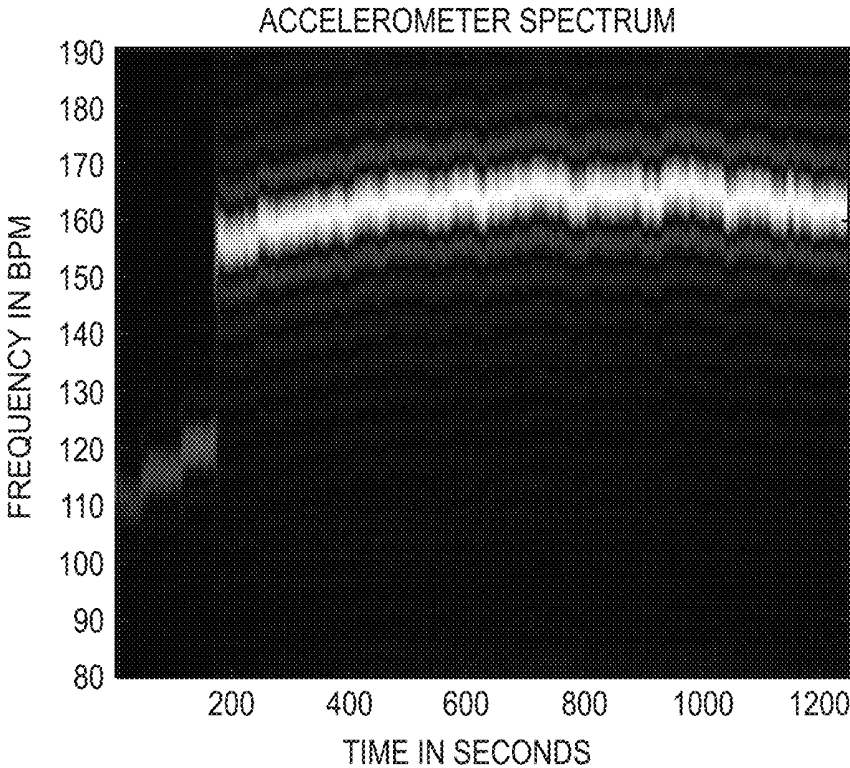
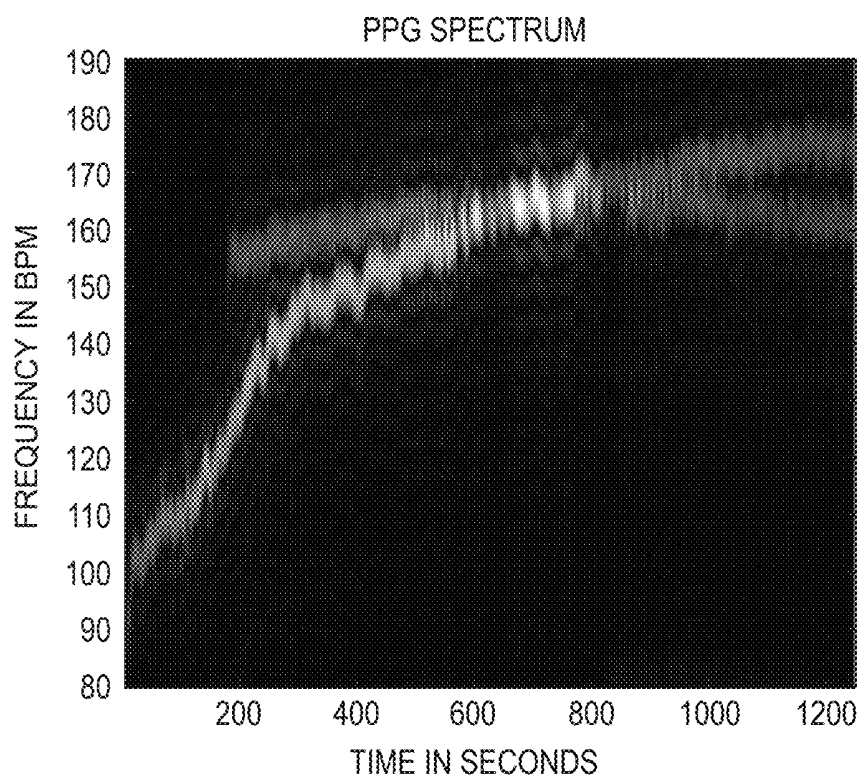


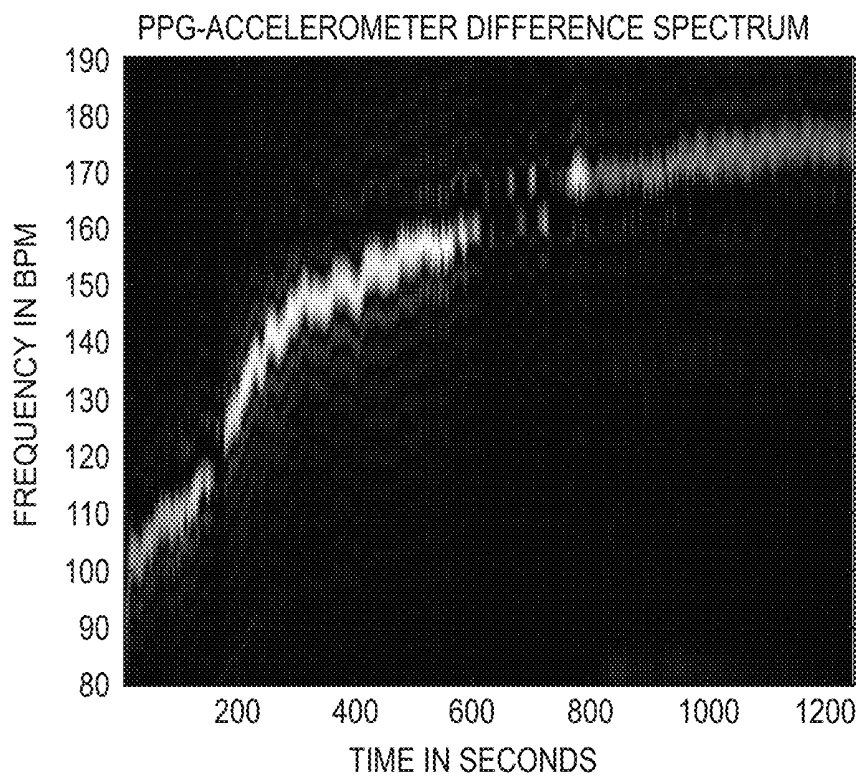
FIG. 5



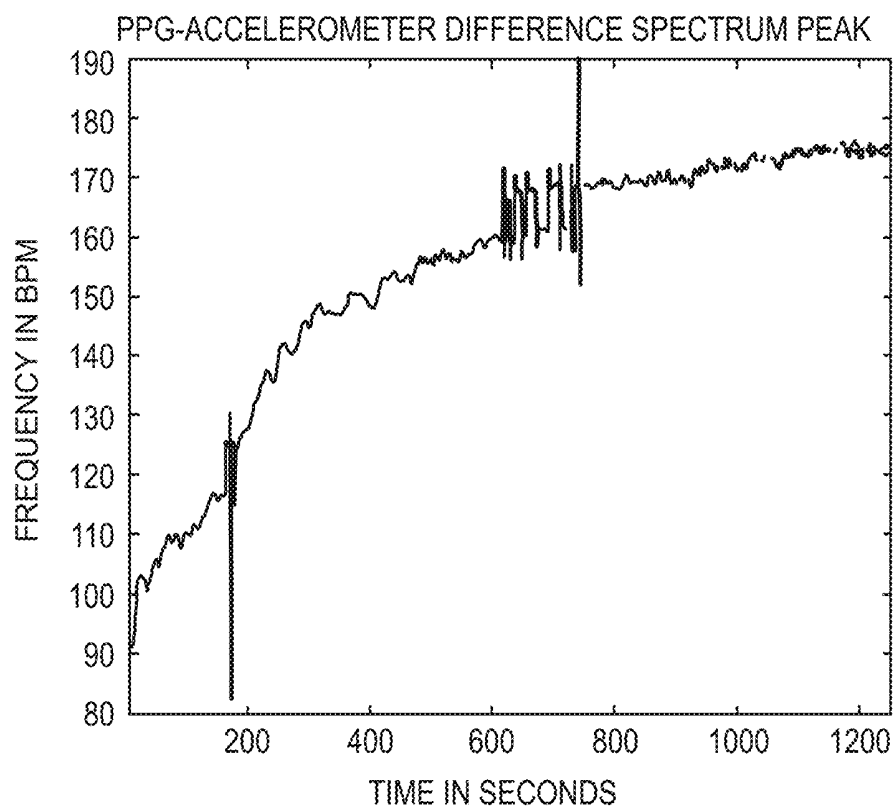
**FIG. 6A**



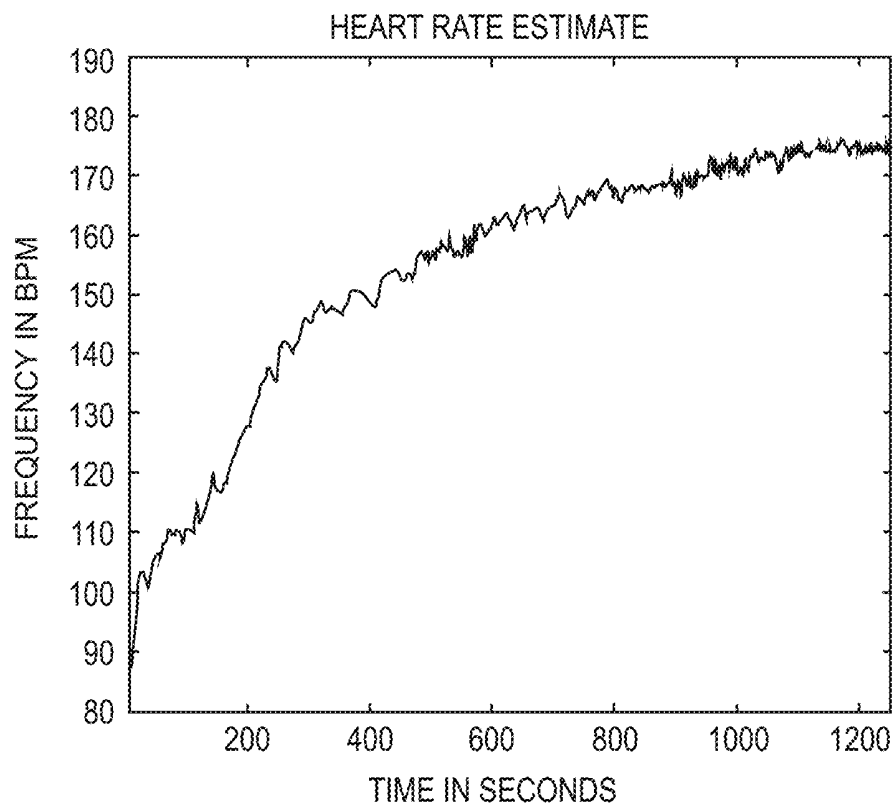
**FIG. 6B**



**FIG. 6C**



**FIG. 6D**



**FIG. 7**

## REDUCTION OF PHYSIOLOGICAL METRIC ERROR DUE TO INERTIAL CADENCE

### RELATED APPLICATIONS

This application is a Continuation Application of U.S. application Ser. No. 16/146,362, filed 28 Sep. 2018, which is a Continuation Application of U.S. application Ser. No. 14/370,689, filed 3 Jul. 2014, now patented as U.S. patent Ser. No. 10/349,844, which National Phase of International Application PCT/US2012/071594 filed 24 Dec. 2012, which in turn claims benefit of U.S. Provisional Application No. 61/586,884, filed 16 Jan. 2012. The disclosures of each of these references are incorporated in their entireties by reference herein.

### BACKGROUND

Personal health monitors provide users with the ability to monitor their overall health and fitness by enabling the user to monitor heart rate or other physiological information during exercise, athletic training, rest, daily life activities, physical therapy, etc. Such devices are becoming increasingly popular as they become smaller and more portable.

A heart rate monitor represents one example of a personal health monitor. A common type of heart rate monitor uses a chest strap that includes surface electrodes to detect muscle action potentials from the heart. Because such surface electrodes provide a relatively noise free signal, the information produced by monitors that use surface electrodes is relatively accurate. However, most users find chest strap monitors uncomfortable and inconvenient.

Another type of monitor uses photoplethysmograph (PPG) sensors disposed in an ear bud. The ear provides an ideal location for a monitor because it is a relatively immobile platform that does not obstruct a person's movement or vision. PPG sensors proximate the ear may have, e.g., access to the inner ear canal and tympanic membrane (for measuring core body temperature), muscle tissue (for monitoring muscle tension), the pinna and earlobe (for monitoring blood gas levels), the region behind the ear (for measuring skin temperature and galvanic skin response), and the internal carotid artery (for measuring cardiopulmonary functioning). The ear is also at or near the point of the body's exposure to environmental breathable toxins of interest (volatile organic compounds, pollution, etc.), noise pollution experienced by the ear, lighting conditions for the eye, etc. Further, as the ear canal is naturally designed for transmitting acoustical energy, the ear provides a good location for monitoring internal sounds, such as the heartbeat, breathing rate, mouth motion, etc.

PPG sensors measure the relative blood flow using an infrared or other light source that projects light that is ultimately transmitted through or reflected off tissue, and is subsequently detected by a photodetector and quantified. For example, higher blood flow rates result in less light being absorbed, which ultimately increases the intensity of the light that reaches the photodetector. By processing the signal output by the photodetector, a monitor using PPG sensors may measure the blood volume pulse (the phasic change in blood volume with each heartbeat), the heart rate, heart rate variability, and other physiological information.

PPG sensors are generally small and may be packaged such that they do not encounter the comfort and/or convenience issues associated with other conventional health monitors. However, PPG sensors are also highly sensitive to noise, and thus are more prone to accuracy problems. For

example, a motion component of a user, e.g., a step rate of a jogger, is often as strong as or stronger than a heart rate component, which may corrupt a heart rate measurement. U.S. Pat. No. 7,144,375, which discloses using an accelerometer as a motion reference for identifying the potential step rate component(s) of a PPG sensor output, provides one possible solution to this problem. When the step rate is close to the heart rate, the '375 patent teaches spectrally transforming the step rate and heart rate waveforms, e.g., over a window of samples, respectively provided by the step rate and heart rate sensors to create a step rate spectrum and a heart rate spectrum. If the spectral transform operation uses a 6 s window, the average latency incurred for the transform operation is 3 s. After performing the spectral transform, the '375 patent spectrally subtracts the heart rate and step rate spectrums. The '375 patent further keeps a history of the top ten peaks from the output of the spectral subtraction to perform various statistical analyses in order to achieve the desired accuracy before making a decision regarding whether there is a cross-over between the heart rate and the step rate, and before making a decision regarding which spectral peak corresponds to the heart rate. Thus, the post-transform operations implemented by the '375 patent incur an additional processing latency, e.g., of ten seconds, which is undesirable. Thus, there remains a need for alternative solutions that provide an accurate heart rate with less latency when the step rate is close to the heart rate.

### SUMMARY

The solution disclosed herein removes a step rate component from a measured heart rate by using one or more filtering techniques when the step rate is close to the heart rate. In general, a difference between the step rate and the heart rate is determined, and the step rate is filtered from the heart rate based on a function of the difference.

In one exemplary embodiment a step rate processor computes a step rate of a user based on a waveform provided by a step rate sensor, and a heart rate processor computes a first heart rate of the user based on a waveform provided by a heart rate sensor. A noise processor then computes a difference between the step rate and the heart rate, computes a second heart rate of the user as a function of the difference, and outputs the second heart rate. For example, the noise processor may filter the heart rate as a function of the difference.

More broadly, an exemplary physiological monitor comprises an inertial sensor, an inertial processor, a physiological sensor, a physiological processor, and a noise processor. The inertial processor computes an inertial cadence of a user based on an inertial waveform provided by the inertial sensor. The physiological processor computes a first physiological metric of the user based on a physiological waveform provided by the physiological sensor. The noise processor computes a difference between the inertial cadence and the first physiological metric, computes a second physiological metric as a function of the difference, and outputs the second physiological metric.

An exemplary method reduces noise in data output by a physiological monitor. To that end, the method includes computing an inertial cadence of a user based on an inertial waveform provided by an inertial sensor in the physiological monitor, and computing a first physiological metric of the user based on a physiological waveform provided by a physiological sensor in the physiological monitor. Subsequently, the method computes a difference between the inertial cadence and the first physiological metric, computes

a second physiological metric as a function of the difference, and outputs the second physiological metric.

Because the solution disclosed herein processes only the current spectrally transformed data, e.g., the current step rate and heart rate spectrums, the present invention essentially eliminates the post-transform latency incurred by the '375 patent. Thus, the solution disclosed herein provides sufficient accuracy without the undesirable latency associated with the prior art.

#### BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 shows an exemplary heart rate monitor disposed in an ear bud.

FIG. 2 shows a block diagram of an exemplary physiological monitoring system disposed in a housing.

FIG. 3 shows an exemplary process for removing an inertial cadence from a physiological metric.

FIG. 4 shows another exemplary process for removing the inertial cadence from the physiological metric.

FIG. 5 shows an exemplary block diagram for the physiological processor of FIG. 2.

FIGS. 6A-6D show exemplary results of simulated step rate and heart rate spectrums, and the resulting difference spectrums and estimated heart rates.

FIG. 7 show an exemplary result of a simulation used to estimate a heart rate according to the solution disclosed herein.

#### DETAILED DESCRIPTION

The techniques disclosed herein improve the accuracy of the results achieved when processing data, e.g., heart rate data, provided by a physiological sensor. FIG. 1 shows an exemplary monitoring system 12 disposed in an ear bud 10. The ear bud 10 may comprise a wireless or wired ear bud that communicatively couples to a remote device, e.g., a music player, a smart phone, a personal data assistant, etc. The monitoring system 12 monitors the heart rate and/or other physiological metrics, and outputs such physiological information to the user and/or to other processing functions. While the monitoring system 12 disclosed herein is presented as being part of an ear bud 10, it will be appreciated that monitoring system 12 may be disposed into any device that secures to the body of a user, e.g., a device that secures to an ear, finger, toe, limb, wrist, nose, etc. In some embodiments, the device may comprise a patch, e.g., a bandage, designed to attach the system 12 to any desired location on the user's body.

FIG. 2 shows a block diagram of an exemplary monitoring system 12 according to one exemplary embodiment. System 12 comprises a processor 100 coupled to one or more physiological sensors 20 and one or more inertial sensors 30, an output interface 40, and memory 50. Physiological sensor(s) 20 produce a physiological waveform responsive to the physiological state of the user. Inertial sensor(s) 30 produce an inertial waveform responsive to the movement of the user. An exemplary inertial sensor includes, but is not limited to, an accelerometer, an optical emitter/detector pair, an optical detector, a CCD camera, a piezoelectric sensor, a thermal sensor, or any type of sensor capable of capturing motion information. Exemplary optical emitters comprise one or more light emitting diodes, laser diodes, organic light-emitting diodes, miniature light emitters, electromagnetic emitters, etc. It will be appreciated that the sensors disclosed herein are not limited to optical wavelengths of the electromagnetic spectrum. In some

embodiments, emitters and/or detectors configured for shorter or longer wavelengths may be used to accommodate shorter or longer wavelengths in the electromagnetic spectrum. The optical detector may comprise a photodetector, an electromagnetic detector, a photodiode, etc. Processor 100 processes the physiological and inertial waveforms using the technique(s) disclosed herein, along with information stored in memory 50, to remove an inertial cadence from a heart rate and/or one or more physiological metrics. Output interface 40 outputs the determined physiological metric(s). It will be appreciated that output interface may comprise a transceiver for transmitting the data output by the processor 100 to a remote device. Alternatively or additionally, the output interface may provide the output data to a user interface, e.g., a display, a database, another processor, and/or a processing function.

While the physiological sensors 20 may comprise any known physiological sensor, the physiological sensor(s) 20 in exemplary embodiments comprise photoplethysmograph (PPG) sensors that generate an electrical physiological waveform responsive to detected light intensity. PPG sensors comprise light intensity sensors that generally rely on optical coupling of light into the blood vessels. As used herein, the term "optical coupling" refers to the interaction or communication between excitation light entering a region and the region itself. For example, one form of optical coupling may be the interaction between excitation light generated from within a light-guiding ear bud 10 and the blood vessels of the ear. Light guiding ear buds are described in co-pending U.S. Patent Application Publication No. 2010/0217102, which is incorporated herein by reference in its entirety. In one embodiment, the interaction between the excitation light and the blood vessels may involve excitation light entering the ear region and scattering from a blood vessel in the ear such that the intensity of the scattered light is proportional to blood flow within the blood vessel. Another form of optical coupling may result from the interaction between the excitation light generated by an optical emitter within the ear bud 10 and the light-guiding region of the ear bud 10.

Processor 100 comprises an inertial processor 110, physiological processor 120, and noise processor 140. Inertial processor 110 determines an inertial cadence I, e.g., a step rate, from the inertial waveform using any known means. The determined inertial cadence may include the true inertial cadence as well as one or more harmonics of the true inertial cadence, e.g., the  $\frac{1}{2}x$ ,  $\frac{1}{2}x$ , and  $2x$  harmonics of the true inertial cadence. For example, the inertial processor may spectrally transform the inertial waveform to generate an inertial spectrum, and set the inertial cadence to the frequency of the largest peak of the inertial spectrum. It will be appreciated that other methods may alternatively be used to determine the inertial cadence. Physiological processor 120 determines one or more physiological metrics H, e.g., a heart rate, from the physiological waveform, as discussed further herein. The determined physiological metric may also refer to a physiological assessment computed from one or more physiological metrics. Noise processor 140 filters the determined metric(s) to remove the inertial cadence, and therefore, to produce a revised physiological metric H having an improved accuracy.

For simplicity, the following describes the processor 100 in terms of a noise processor 140 that determines a heart rate by removing a step rate by using one or more filtering techniques when the step rate determined by the inertial processor 110 is close to the heart rate determined by the physiological processor 120. In general, a difference

between the step rate and the heart rate is determined, and the step rate is filtered from the heart rate based on a function of the difference. It should be noted, however, that the physiological processor **120**, and thus processor **100**, may alternatively or additionally determine other physiological metrics, e.g., a respiration rate, a heart rate variability (HRV), a pulse pressure, a systolic blood pressure, a diastolic blood pressure, a step rate, an oxygen uptake (VO<sub>2</sub>), a maximal oxygen uptake (VO<sub>2</sub> max), calories burned, trauma, cardiac output and/or blood analyte levels including percentage of hemoglobin binding sites occupied by oxygen (SPO<sub>2</sub>), percentage of methemoglobins, a percentage of carbonyl hemoglobin, and/or a glucose level. Alternatively or additionally, processor **100** may determine and filter one or more physiological assessments, e.g., a ventilatory threshold, lactate threshold, cardiopulmonary status, neurological status, aerobic capacity (VO<sub>2</sub> max), and/or overall health or fitness. Further, it will be appreciated that the processor **100** may additionally or alternatively remove other inertial cadences, e.g., rhythmic head movements, body movements (e.g., arm movements, weight lifting, etc.), etc., from the heart rate.

FIG. 3 shows an exemplary method **200** that may be implemented by processor **100** to compute an output heart rate. After processor **100** receives the inertial and physiological waveforms from the sensor(s) **20** and **30** (block **210**), the inertial processor **110** determines a step rate *I* (block **220**), and the physiological processor **120** determines a first estimate *H* of the heart rate (block **230**). The noise processor **140** computes a revised estimate  $\hat{H}$  of the heart rate as a function of the difference between the step rate and the heart rate (block **240**), and subsequently outputs the revised heart rate estimate *H* to output interface **40** (block **290**). It will be appreciated that processor **100** may also store the revised heart rate estimate  $\hat{H}$  in memory **50**, e.g., so that it may be used in subsequent calculations.

FIG. 4 shows an exemplary method **240** for computing the revised heart rate  $\hat{H}$  as a function of the difference between the step rate and the first heart rate. The exemplary method **240** of FIG. 4 includes multiple levels, where a first level includes block **242**, a second level includes blocks **244** to **248**, a third level includes block **250**, a fourth level includes block **260**, a fifth level includes block **270**, and a sixth level includes blocks **280** to **284**. It will be appreciated, however, that some exemplary embodiments may implement one or more of these levels, as described further herein, and/or that the present invention does not require each level to be implemented in the order shown in FIG. 4.

The first level (blocks **242** and **243**) comprises an initialization level, where the heart rate processor **120** and/or noise processor **140** initializes and/or determines one or more variables useful for determining the output heart rate based on pre-determined values, values stored in memory, and measured information (blocks **242** and **243**), e.g., an instantaneous heart rate  $H_{inst}$ , a filtered heart rate  $H_{filt}$ , a lock count  $C_{lk}$ , a second (or output) heart rate *H*, etc. For example, to determine  $H_{inst}$ , heart rate processor **120** comprises a spectral transformer **122** (FIG. 5) that spectrally transforms the physiological waveform output by the heart rate sensor(s) **20** to generate a physiological spectrum. For example, the spectral transformer **122** may transform the heart rate waveform over a window of samples, e.g., 6 s. It will be appreciated that such a transform operation incurs a latency half the length of the window, e.g., 3 s. The heart rate processor **120** initializes the first heart rate by identifying the frequency  $F_1$  of the spectral peak of the physiological spectrum having the largest amplitude as an initial instan-

aneous heart rate  $H_{inst}$ , where the initial instantaneous heart rate corresponds to the first heart rate (block **243**). The heart rate processor **1120** may further identify the frequency  $F_2$  of the spectral peak of the physiological spectrum having the next largest amplitude. Alternatively or additionally, a filtered heart rate may be initialized, e.g., by retrieving a previously determined filtered heart rate from memory **50** or by setting  $H_{filt}$  to an empirically determined value, e.g., 83. Further,  $C_{lk}$  may be initialized by setting it equal to zero, where  $C_{lk}$  represents, e.g., the number of successive frames where the difference between the heart rate and the step rate satisfies a predetermined threshold. Noise processor **140** may also determine an initial second (or output) heart rate *H*, where the second heart rate represents the heart rate ultimately output by the noise processor **140**. In some cases, the processor **140** may set the initial *H* to an empirically determined value, e.g., 83. Alternatively or additionally, the noise processor **140** may initialize *H* to the  $\hat{H}$  value from a previous frame after the first iteration through the process **240**.

The second level (blocks **244** to **248**) determines an instantaneous heart rate as a function of a difference between an initial instantaneous heart rate and the step rate, and particularly addresses the scenario when the results of spectral subtraction wipe out the primary spectral peak typically used to determine the instantaneous heart rate. More particularly, the second level determines whether the step rate *I* provided by inertial processor **110** is close to the initial instantaneous heart rate  $H_{inst}$  by determining whether the initial instantaneous heart rate is within a crossover window (block **244**), and adjusts the instantaneous heart rate based on that determination. For example, noise processor **140** may determine whether the initial  $H_{inst}$  is within the crossover window by determining whether the difference between the step rate *I* and the initial  $H_{inst}$  is less than or equal to a threshold  $T_w$ , e.g.,  $T_w=8$ . In one embodiment, noise processor **140** adjusts the initial  $H_{inst}$  only when the initial  $H_{inst}$  is within the crossover window, where the adjustment is based on a weighted average of the frequencies of two or more of the spectral peaks provided by the spectral transformer **122** (block **248**). For example, noise processor **140** may compute a weight *w* according to:

$$w = \frac{M_1}{M_1 + M_2}, \quad (1)$$

where  $M_1$  represents the magnitude of the largest spectral peak of the physiological spectrum and  $M_2$  represents the magnitude of a second spectral peak of the physiological spectrum, e.g., the next largest spectral peak. The noise processor **140** then adjusts the instantaneous heart rate by computing a weighted average of the frequencies of the two spectral peaks, e.g., according to:

$$H_{inst} = wF_1 + (1-w)F_2, \quad (2)$$

where  $F_1$  represents the frequency of the largest spectral peak (and corresponds to the initial instantaneous heart rate), and  $F_2$  represents the frequency of the second spectral peak.

In some embodiments, the second level may also optionally determine whether the frequency  $F_2$  of the second spectral peak of the physiological spectrum is also in the crossover window and whether the step rate *I* is between the initial  $H_{inst}$  and  $F_2$  (block **246**) when the condition of block **244** is satisfied. For example, the noise processor **140** may determine whether  $F_2$  is within the crossover window by

determining whether the difference between  $F_2$  and  $I$  is less than or equal to a threshold, e.g., 8. Further, the noise processor **140** may determine whether  $I$  is between  $F_2$  and the initial  $H_{inst}$  by determining whether  $\text{sign}(H_{inst}-I) \neq \text{sign}(F_2-I)$ . In any event, for this example, noise processor **140** executes the operation of block **248** only when the conditions of blocks **244** and **246** are both satisfied.

The third level (block **250**) filters the instantaneous heart rate using rate limits. More particularly, the third level computes a revised filtered heart rate  $\hat{H}_{filt}$  as a function of a current filtered heart rate  $H_{filt}$ , the instantaneous heart rate  $H_{inst}$  output by the second level, and a rate limit  $\Delta_r$ . For this embodiment, the heart rate processor **120** may further include a filter **124**, as shown in FIG. **5**. In general, after comparing  $H_{inst}$  to  $H_{filt}$  which may be retrieved from memory **50** or provided by initialization block **242**, filter **124** computes the revised filtered heart rate estimate  $\hat{H}_{filt}$  as a function of  $H_{filt}$ ,  $H_{inst}$ , and the rate limit  $\Delta_r$  based on the comparison, where  $\Delta_r$  may also be retrieved from memory **50**. In one exemplary embodiment, when  $H_{inst} \geq H_{filt}$ , filter **124** computes the revised filter estimate  $\hat{H}_{filt}$  according to:

$$\hat{H}_{filt} = H_{filt} + \min(\Delta_r, H_{inst} - H_{filt}) \quad (3)$$

where  $\Delta_{r+}$  represents an increasing rate limit. However, when  $H_{inst} < H_{filt}$ , filter **124** computes the revised filter estimate  $\hat{H}_{filt}$  according to:

$$\hat{H}_{filt} = H_{filt} + \max(\Delta_r, H_{inst} - H_{filt}), \quad (4)$$

where  $\Delta_{r-}$  represents a decreasing rate limit. As used herein, the rate limit represents a limit to the rate of change for the heart rate. For example, the rate limit may represent the rate of change in beats per minute (BPM) that the heart rate may experience in a 1 second frame period. Such a rate limit may be determined based on empirical evidence, and is generally predetermined. It will be appreciated that the rate limit may be expressed as the rate of change experienced for any length frame period, where for example, the rate limit in BPM/s is multiplied by the length of the frame period (in seconds). Additional details for the implementation of block **250** may be found in U.S. Provisional Application Ser. No. 61/586,874 filed concurrently herewith and titled "Physiological Metric Estimation Rise and Fall Limiting," which is incorporated by reference herein in its entirety.

In the fourth level (block **260**), the noise processor **140** biases the filtered heart rate toward the step rate to minimize wandering due to blindness during a crossover. To that end, the noise processor **140** further adjusts the revised filter estimate  $\hat{H}_{filt}$  as a function of the differenced between the step rate and  $\hat{H}_{filt}$ . For example, the noise processor **140** may determine whether the revised filter estimate  $\hat{H}_{filt}$  output by block **250** is within a crossover window by comparing the difference between  $\hat{H}_{filt}$  and  $I$  to a threshold, e.g.,  $\text{abs}(I - \hat{H}_{filt}) \leq 8$ . If  $\hat{H}_{filt}$  is within the crossover window, noise processor **140** may further adjust  $\hat{H}_{filt}$  as a function of the difference between  $\hat{H}_{filt}$  and  $I$ . For example, the noise processor **140** may further adjust  $\hat{H}_{filt}$  according to:

$$\hat{H}_{filt} = \hat{H}_{filt} + 0.5 \text{sign}(I - \hat{H}_{filt}). \quad (5)$$

In the fifth level (block **270**), the noise processor **140** counts the number of successive frames where the heart rate is within a crossover window. To that end, the noise processor updates a lock count  $C_{lk}$  as a function of the difference between  $I$  and  $\hat{H}_{filt}$  output by block **260**, where  $C_{lk}$  represents, e.g., the number of successive frames where the difference between  $\hat{H}_{filt}$  and  $I$  satisfy a threshold requirement. For example, when  $C_{lk}$  is compared to 0 for equality (e.g.,  $C_{lk} = 0$ ), and when  $\text{abs}(H_{filt} - I) < 6$ , noise processor **140**

may set  $C_{lk} = 1$ . However, when  $C_{lk} > 0$ , and when  $\text{abs}(H_{filt} - I) > 6$ , noise processor **140** may set  $C_{lk} = 0$ , and when  $C_{lk} > 0$ , and when  $\text{abs}(H_{filt} - I) \leq 6$ , noise processor **140** may increment  $C_{lk}$ , e.g., set  $C_{lk} = C_{lk} + 1$ .

In the sixth level (blocks **280** to **284**), the noise processor **140** filters the oscillations of the instantaneous heart rate that occur during a crossover sustained for a number of successive frames. For example, the  $\hat{H}_{filt}$  output by block **260** is further filtered responsive to a comparison between the lock count and a threshold  $T_c$  to generate the second (or output) heart rate  $\hat{H}$  to be output to the output interface **40**. For example, if  $C_{lk} > T_c$ , the output heart rate  $\hat{H}$  may be determined as a first function of a previously determined (or initialized) output heart rate  $H$  and  $\hat{H}_{filt}$ , e.g., according to  $f_1(H, \hat{H}_{filt})$  (block **282**). In one exemplary embodiment, the first function may comprise:

$$\hat{H} = f_1(H, \hat{H}_{filt}) = H + (\hat{H}_{filt} - H)/4 \quad (6)$$

where  $H$  represents an initialized second heart rate from block **242**, or a previously determined second (or output) heart rate that may, e.g., be retrieved from memory **50**. If, however,  $C_{lk} \leq T_c$ , the output heart rate  $H$  may be determined as a second function of the previously determined (or initialized) output heart rate  $H$  and  $\hat{H}_{filt}$ , e.g., according to  $f_2(H, \hat{H}_{filt})$  (block **284**). In one exemplary embodiment, the second function may comprise:

$$\hat{H} = f_2(H, \hat{H}_{filt}) = H + (\hat{H}_{filt} - H)/2. \quad (7)$$

It will be appreciated that not all of the six levels of FIG. **4** are required to determine  $\hat{H}$  output to the output interface **40**. For example, exemplary embodiments may compute the output heart rate using:

the second and third levels, where the output heart rate comprises the filtered heart rate  $\hat{H}_{filt}$  output by filter **124** of FIG. **5** and block **250** of FIG. **4**.

the third and fourth levels, where the output heart rate comprises the filtered heart rate  $\hat{H}_{filt}$  output by block **260** of FIG. **4**, and the input heart rates comprise e.g., an initial instantaneous heart rate  $H_{inst}$  from block **242** of Figure and an initial filtered heart rate  $H_{filt}$  obtained from block **242** of FIG. **4** or previously determined and retrieved from memory **50**.

the fourth level, where the output heart rate comprises the filtered heart rate  $\hat{H}_{filt}$  output by block **260** of FIG. **4**, and the input heart rate comprises, e.g., an initial filtered heart rate  $H_{filt}$  obtained from block **242** of FIG. **4** or previously determined and retrieved from memory **50**.

the third, fourth, fifth, and sixth levels, where the output heart rate comprises  $\hat{H}$  as output by one of blocks **282** and **284** of FIG. **4**, and the input heart rates comprise an initial instantaneous heart rate  $H_{inst}$  from block **242** of Figure and an initial filtered heart rate  $H_{filt}$  obtained from block **242** of FIG. **4** or previously determined and retrieved from memory **50**.

the fourth, fifth, and sixth levels, where the output heart rate comprises  $\hat{H}$  as output by one of blocks **282** and **284** of FIG. **4**, and the input heart rate comprises an initial filtered heart rate obtained from block **242** of FIG. **4** or previously determined and retrieved from memory **50**.

the fifth and sixth levels, where the output heart rate comprises  $\hat{H}$  as output by one of blocks **282** and **284** of FIG. **4**, and the input heart rates comprise an initial second heart rate  $H$  obtained from block **242** of FIG. **4**

and an initial filtered heart rate obtained from block 242 of FIG. 4 or previously determined and retrieved from memory 50.

the sixth level, where the output heart rate comprises  $\hat{H}$  as output by one of blocks 282 and 284 of FIG. 4, and the input heart rates comprise an initial second heart rate  $H$  obtained from block 242 of FIG. 4 and an initial filtered heart rate obtained from block 242 of FIG. 4 or previously determined and retrieved from memory 50. In this case, the decision diamond 280 not only considers how the lock count compares to the threshold, but it also considers whether the filtered heart rate is within a crossover window, e.g., by determining whether the difference between  $\hat{H}_{fit}$  and  $I$  is less than or equal to a threshold  $T_w$ .

the third, fifth, and sixth levels, where the output heart rate comprises  $\hat{H}$  as output by one of blocks 282 and 284 of FIG. 4, and the input heart rates comprise an initial instantaneous heart rate  $H_{inst}$  from block 242 of Figure and an initial filtered heart rate  $H_{fit}$  obtained from block 242 of FIG. 4 or previously determined and retrieved from memory 50.

It will be appreciated that other combinations not explicitly disclosed herein may also be used to generate the output heart rate.

FIGS. 6A-6D and 7 further illustrate both the impact of a step rate on a measured heart rate, and how the solution disclosed herein addresses this problem. First, FIGS. 6A-6D simulate the problems that occur when the step rate and the heart rate are close together. In particular, FIG. 6A shows a step rate spectrum generated responsive to an accelerometer waveform, while FIG. 6B shows a heart rate spectrum generated responsive to a waveform output by a PPG sensor, and thus includes both step rate and heart rate elements. FIG. 6C shows the spectrum produced when the accelerometer spectrum of FIG. 6A is simply subtracted from the heart rate spectrum of FIG. 6B. As shown in FIG. 6C, when the step rate is close to the heart rate, e.g., between 600 and 800 s, the heart rate is attenuated. Thus, while simple spectral subtraction removes the step rate component, it also corrupts the heart rate measurements subsequently obtained from the heart rate spectrum. In particular, FIG. 6D shows how the peak frequency of the difference spectrum of FIG. 6C has large oscillating errors in the crossover window.

FIG. 7 shows the simulation results when the heart rate is estimated using the techniques disclosed herein, e.g., the techniques of FIG. 4. As shown in FIG. 7, the errors around the crossover regions have nearly disappeared, which results in a more accurate heart rate estimate.

The solution disclosed herein provides an accurate heart rate estimate, as shown for example by the simulated results of FIG. 7, without incurring the latency problems associated with the known prior art solution. In particular, because the solution disclosed herein does not require multiple spectrums or statistics to be buffered in a look-ahead fashion in order to detect the initial crossover, which leads to an undesirable latency, the solution disclosed herein avoids the latency problems associated with the prior art without sacrificing accuracy.

While the present invention is described in terms of PPG sensors, it will be appreciated that sensors 20 may comprise any sensor able to generate a physiological waveform, e.g., an electroencephalogram (EEG) waveform, and electrocardiogram (ECG) waveform, a radio frequency (RF) waveform, an electro-optical physiological waveform, and electro-photoacoustic waveform including a photacoustic

waveform, an electro-mechanical physiological waveform, and/or an electro-nuclear physiological waveform.

The present invention may, of course, be carried out in other ways than those specifically set forth herein without departing from essential characteristics of the invention. The present embodiments are to be considered in all respects as illustrative and not restrictive, and all changes coming within the meaning and equivalency range of the appended claims are intended to be embraced therein.

What is claimed is:

1. A heart rate monitor for sensing a heart rate of a user, the heart rate monitor comprising:

a photoplethysmograph (PPG) sensor configured to: sense light scattered from a body of the user; and provide PPG information responsive to the sensed light;

a step rate sensor configured to: sense motion from the user; and provide motion information responsive to the sensed motion;

at least one processor circuit configured to: determine a step rate of a user from the motion information provided by the step rate sensor; determine an initial instantaneous heart rate from the PPG information provided by the PPG sensor and the motion information provided by the step rate sensor; determine a revised heart rate, responsive to whether the initial instantaneous heart rate is within a first crossover window relative to the step rate, from the initial instantaneous heart rate and the step rate; determine an output heart rate from the revised heart rate; and output the output heart rate.

2. The heart rate monitor of claim 1 wherein the at least one processor circuit determines the revised heart rate by: determining a second instantaneous heart rate, responsive to whether the initial instantaneous heart rate is within the first crossover window relative to the step rate, from the initial instantaneous heart rate and the step rate; and determine the revised heart rate from the second instantaneous heart rate.

3. The heart rate monitor of claim 2: wherein the at least one processor circuit is further configured to compute a difference between the initial instantaneous heart rate and the step rate; and wherein the at least one processor circuit determines the second instantaneous heart rate by determining the second instantaneous heart rate, responsive to whether the difference is less than a threshold, as a function of the initial instantaneous heart rate.

4. The heart rate monitor of claim 3: wherein the at least one processor circuit is further configured to determine a weighted average of frequencies of two or more spectral peaks of the PPG information; and

wherein the at least one processor circuit determines the second instantaneous heart rate by determining the second instantaneous heart rate as a function of the weighted average when the difference is less than the threshold.

5. The heart rate monitor of claim 1 wherein: the initial instantaneous heart rate corresponds to a first frequency of the PPG waveform; the at least one processor circuit is further configured to determine whether a second frequency of the PPG waveform corresponding to a second peak of the PPG waveform is within the first crossover window; and

11

the at least one processor circuit further determines the revised heart rate responsive to whether the second frequency is within the first crossover window.

6. The heart rate monitor of claim 5 wherein the at least one processor circuit further determines the revised heart rate responsive to whether the step rate is between the initial instantaneous heart rate and the second frequency.

7. The heart rate monitor of claim 1 wherein the at least one processor circuit determines the revised heart rate by: determining whether the revised heart rate is within the first crossover window responsive to a comparison of a difference between the revised heart rate and the step rate to a threshold; and

when the revised heart rate is within the first crossover window, adjusting the revised heart rate as a function of the difference between revised heart rate and the step rate to generate an adjusted revised heart rate.

8. The heart rate monitor of claim 1 wherein the at least one processor circuit is further configured to determine a number of successive frames where the revised heart rate is within the first crossover window to determine a lock count.

9. The heart rate monitor of claim 8 wherein the at least one processor circuit determines the output heart rate by: determining the output heart rate as a first function of an initialized second heart rate and the revised heart rate when the lock count exceeds a threshold; and determining the output heart rate as a second function of the initialized second heart rate and the revised heart rate when the lock count does not exceed the threshold; wherein the first function differs from the second function.

10. A method of sensing a heart rate of a user comprising: sensing light scattered from a body of the user using a photoplethysmograph (PPG) sensor, said PPG sensor providing PPG information responsive to the sensed light;

sensing motion from the user using a step rate sensor, said step rate sensor providing motion information responsive to the sensed motion;

determining a step rate of a user from the motion information provided by the step rate sensor;

determining an initial instantaneous heart rate from the PPG information provided by the PPG sensor and the motion information provided by the step rate sensor;

determining a revised heart rate, responsive to whether the initial instantaneous heart rate is within a first crossover window relative to the step rate, from the initial instantaneous heart rate and the step rate;

determining an output heart rate from the revised heart rate; and

outputting the output heart rate to the user.

11. The method of claim 10 wherein the determining the revised heart rate comprises:

determining a second instantaneous heart rate, responsive to whether the initial instantaneous heart rate is within

12

the first crossover window relative to the step rate, from the initial instantaneous heart rate and the step rate; and determine the revised heart rate from the second instantaneous heart.

12. The method of claim 11 further comprising computing a difference between the initial instantaneous heart rate and the step rate, wherein the determining the second instantaneous heart rate comprises determining the second instantaneous heart rate, responsive to whether the difference is less than a threshold, as a function of the initial instantaneous heart rate.

13. The method of claim 12 further comprising determining a weighted average of frequencies of two or more spectral peaks of the PPG information, wherein the determining the second instantaneous heart rate comprises determining the second instantaneous heart rate as a function of the weighted average when the difference is less than the threshold.

14. The method of claim 10 wherein the initial instantaneous heart rate corresponds to a first frequency of the PPG waveform, the method further comprising:

determining whether a second frequency of the PPG waveform corresponding to a second peak of the PPG waveform is within the first crossover window; and determining the revised heart rate responsive to whether the second frequency is within the first crossover window.

15. The method of claim 14 wherein the determining the revised heart rate comprises determining the revised heart rate responsive to whether the step rate is between the initial instantaneous heart rate and the second frequency.

16. The method of claim 10 wherein the determining the revised heart rate comprises:

determining whether the revised heart rate is within the first crossover window responsive to a comparison of a difference between the revised heart rate and the step rate to a threshold; and

when the revised heart rate is within the first crossover window, adjusting the revised heart rate as a function of the difference between revised heart rate and the step rate to generate an adjusted revised heart rate.

17. The method of claim 10 further comprising determining a number of successive frames where the revised heart rate is within the first crossover window to determine a lock count.

18. The method of claim 17 wherein the determining the output heart rate comprises:

determining the output heart rate as a first function of an initialized second heart rate and the revised heart rate when the lock count exceeds a threshold; and

determining the output heart rate as a second function of the initialized second heart rate and the revised heart rate when the lock count does not exceed the threshold; wherein the first function differs from the second function.

\* \* \* \* \*

专利名称(译)	减少惯性节奏导致的生理指标误差		
公开(公告)号	<a href="#">US10631740</a>	公开(公告)日	2020-04-28
申请号	US16/694402	申请日	2019-11-25
[标]申请(专利权)人(译)	瓦倫賽爾公司		
申请(专利权)人(译)	VALENCELL INC.		
当前申请(专利权)人(译)	VALENCELL INC.		
[标]发明人	ROMESBURG ERIC DOUGLAS		
发明人	ROMESBURG, ERIC DOUGLAS		
IPC分类号	A61B5/024 A61B5/00 A61B5/11		
CPC分类号	A61B5/721 A61B5/024 A61B5/112 A61B5/6817 A61B2562/0219 A63B2230/04 A63B2220/17 A61B5/0205 A61B5/021 A61B5/02405 A61B5/0816 A61B5/14542 A61B5/7207 A61B5/725 A61B2562/02		
优先权	PCT/US2012/071594 2012-12-24 WO 61/586884 2012-01-16 US		
其他公开文献	US20200085314A1		
外部链接	<a href="#">Espacenet</a>		

摘要(译)

当步速接近心率时，本文公开的心率监测器通过使用一种或多种滤波技术从测量的心率中去除步率分量。通常，确定步速和心率之间的差，并且基于该差的函数从心率中过滤步率。

