



US009521978B2

(12) **United States Patent**
Freeman et al.

(10) **Patent No.:** **US 9,521,978 B2**
(45) **Date of Patent:** ***Dec. 20, 2016**

(54) **USING CHEST VELOCITY TO PROCESS
PHYSIOLOGICAL SIGNALS TO REMOVE
CHEST COMPRESSION ARTIFACTS**

(58) **Field of Classification Search**

None

See application file for complete search history.

(71) Applicant: **ZOLL Medical Corporation,**
Chelmsford, MA (US)

(56) **References Cited**

U.S. PATENT DOCUMENTS

(72) Inventors: **Gary A. Freeman**, Waltham, MA (US);
Qing Tan, Somerville, MA (US);
Frederick J. Geheb, Danvers, MA
(US)

3,865,101 A	2/1975	Saper et al.
4,059,099 A	11/1977	Davis
4,088,138 A	5/1978	Diack et al.
4,198,963 A	4/1980	Barkalow et al.
RE30,372 E	8/1980	Mirowski et al.
4,296,755 A	10/1981	Judell

(Continued)

(73) Assignee: **ZOLL Medical Corporation,**
Chelmsford, MA (US)

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

This patent is subject to a terminal dis-
claimer.

FOREIGN PATENT DOCUMENTS

EP	1057451	12/2000
EP	1079310	2/2001

(Continued)

(21) Appl. No.: **14/512,167**

OTHER PUBLICATIONS

(22) Filed: **Oct. 10, 2014**

Aase et al., "Compression Depth Estimation for CPR Quality
Assessment Using DSP on Accelerometer Signals," IEEE Transac-
tions on Biomedical Engineering, vol. 49, No. 3, Mar. 2002.

(65) **Prior Publication Data**

US 2015/0031972 A1 Jan. 29, 2015

(Continued)

Related U.S. Application Data

(63) Continuation of application No. 13/872,033, filed on
Apr. 26, 2013, now Pat. No. 8,862,228, which is a
(Continued)

Primary Examiner — Brian T Gedeon

(74) *Attorney, Agent, or Firm* — Fish & Richardson P.C.

(51) **Int. Cl.**

A61N 1/39 (2006.01)

A61B 5/04 (2006.01)

(Continued)

(57) **ABSTRACT**

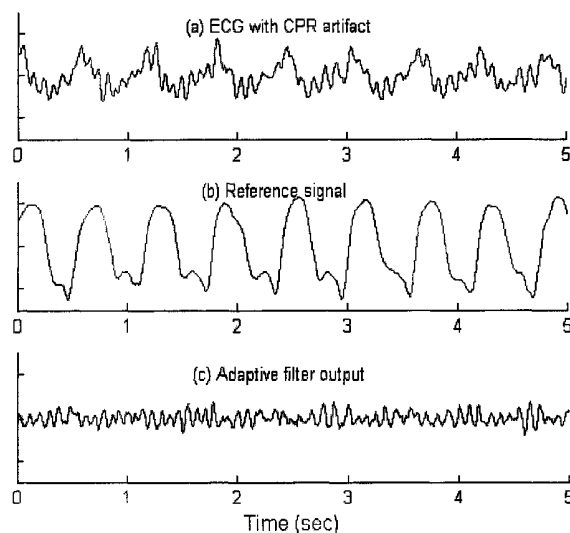
A method of analyzing a physiological (e.g., an ECG) signal
during application of chest compressions. The method
includes acquiring a physiological signal during application
of chest compressions; acquiring the output of a sensor from
which information on the velocity of chest compressions can
be determined; and using the information on the velocity to
reduce at least one signal artifact in the physiological signal
resulting from the chest compressions.

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

CPC **A61B 5/721** (2013.01); **A61B 5/0205**
(2013.01); **A61B 5/046** (2013.01);

(Continued)

31 Claims, 5 Drawing Sheets



Related U.S. Application Data

continuation of application No. 10/786,359, filed on Feb. 24, 2004, now abandoned, which is a continuation-in-part of application No. 10/704,366, filed on Nov. 6, 2003, now Pat. No. 7,220,235.

(51) Int. Cl.

A61B 5/00 (2006.01)
A61B 5/046 (2006.01)
A61H 31/00 (2006.01)
A61B 5/0205 (2006.01)
A61B 5/11 (2006.01)
A61B 5/1455 (2006.01)

(52) U.S. Cl.

CPC *A61B 5/04017* (2013.01); *A61B 5/11* (2013.01); *A61B 5/14551* (2013.01); *A61B 5/4836* (2013.01); *A61B 5/4848* (2013.01); *A61B 5/7207* (2013.01); *A61H 31/005* (2013.01); *A61H 31/006* (2013.01); *A61H 31/007* (2013.01); *A61N 1/3987* (2013.01); *A61H 2201/5058* (2013.01); *A61H 2201/5079* (2013.01); *A61H 2201/5084* (2013.01); *A61H 2230/04* (2013.01); *A61N 1/3925* (2013.01)

(56)**References Cited**

U.S. PATENT DOCUMENTS

4,355,634 A 10/1982 Kanter
 4,588,383 A 5/1986 Parker et al.
 4,610,254 A 9/1986 Morgan et al.
 4,619,265 A 10/1986 Morgan et al.
 4,680,708 A 7/1987 Ambos et al.
 4,757,821 A 7/1988 Snyder
 4,781,200 A 11/1988 Baker
 4,928,674 A 5/1990 Halperin et al.
 5,077,667 A 12/1991 Brown et al.
 5,081,993 A 1/1992 Kitney et al.
 5,092,341 A 3/1992 Kelen
 5,109,862 A 5/1992 Kelen et al.
 5,193,537 A 3/1993 Freeman
 5,211,179 A 5/1993 Haberl et al.
 5,247,945 A 9/1993 Heinze et al.
 5,285,792 A 2/1994 Sjoquist et al.
 5,330,526 A 7/1994 Fincke et al.
 5,348,008 A 9/1994 Bornn et al.
 RE34,800 E 11/1994 Hutchins
 5,391,187 A 2/1995 Freeman
 5,409,010 A 4/1995 Beach et al.
 5,439,483 A 8/1995 Duong-Van
 5,454,779 A 10/1995 Lurie et al.
 5,466,244 A 11/1995 Morgan
 5,471,991 A 12/1995 Shinnar
 5,472,453 A 12/1995 Alt et al.
 5,474,574 A 12/1995 Payne et al.
 5,496,257 A 3/1996 Kelly
 5,507,778 A 4/1996 Freeman
 5,511,553 A 4/1996 Segalowitz
 5,562,710 A 10/1996 Olsen et al.
 5,589,639 A 12/1996 D'Antonio et al.
 5,591,213 A 1/1997 Morgan
 5,611,815 A 3/1997 Cole et al.
 5,617,853 A 4/1997 Morgan
 5,619,265 A 4/1997 Suzuki et al.
 5,626,140 A 5/1997 Feldman et al.
 5,645,571 A 7/1997 Olson et al.
 5,662,690 A 9/1997 Cole et al.
 5,683,424 A 11/1997 Brown et al.
 5,700,281 A 12/1997 Brewer et al.
 5,735,879 A 4/1998 Gliner et al.
 5,755,671 A 5/1998 Albrecht et al.
 5,772,604 A 6/1998 Langberg et al.
 5,782,888 A 7/1998 Sun et al.

5,792,190 A 8/1998 Olson et al.
 5,853,364 A 12/1998 Baker
 5,957,856 A 9/1999 Weil et al.
 5,967,995 A 10/1999 Shusterman et al.
 6,021,345 A 2/2000 Karagueuzian
 6,021,349 A 2/2000 Arand et al.
 6,115,627 A 9/2000 Street
 6,125,299 A 9/2000 Groenke et al.
 6,155,257 A 12/2000 Lurie et al.
 6,171,257 B1 1/2001 Weil et al.
 6,174,295 B1 1/2001 Cantrell et al.
 6,178,357 B1 1/2001 Gliner et al.
 6,188,928 B1 2/2001 Noren et al.
 6,224,562 B1 5/2001 Lurie et al.
 6,246,907 B1 6/2001 Lin et al.
 6,263,238 B1 7/2001 Brewer et al.
 6,289,243 B1 9/2001 Lin et al.
 6,306,107 B1 10/2001 Myklebust et al.
 6,308,094 B1 10/2001 Shusterman et al.
 6,309,695 B1 10/2001 Singh
 6,334,070 B1 12/2001 Nova et al.
 6,351,671 B1 2/2002 Myklebust et al.
 6,360,125 B1 3/2002 Weil et al.
 6,390,996 B1 5/2002 Halperin et al.
 6,393,316 B1 5/2002 Gillberg et al.
 6,397,104 B1 5/2002 Miller et al.
 6,418,342 B1 7/2002 Owen et al.
 6,427,685 B1 8/2002 Ray, II
 6,438,419 B1 8/2002 Callaway et al.
 6,480,734 B1 11/2002 Zhang et al.
 6,496,731 B1 12/2002 Lovett
 6,572,547 B2 6/2003 Miller et al.
 6,575,914 B2 6/2003 Rock et al.
 6,597,943 B2 7/2003 Taha et al.
 6,622,042 B1 9/2003 Thacker
 6,658,290 B1 12/2003 Lin et al.
 6,671,545 B2 12/2003 Fincke
 6,687,540 B2 2/2004 Marcovecchio
 6,807,442 B1 10/2004 Myklebust et al.
 6,827,695 B2 12/2004 Palazzolo et al.
 7,013,176 B2 3/2006 Ding et al.
 7,032,596 B2 4/2006 Thompson et al.
 7,220,235 B2 5/2007 Geheb et al.
 7,310,553 B2 12/2007 Freeman
 7,463,922 B1 12/2008 Snyder et al.
 7,488,293 B2 2/2009 Marcovecchio et al.
 7,565,194 B2 7/2009 Tan et al.
 7,708,683 B2 5/2010 Hadley
 7,831,299 B2 11/2010 Tan et al.
 8,165,671 B2 4/2012 Freeman et al.
 2001/0047140 A1 11/2001 Freeman
 2002/0026131 A1 2/2002 Halperin
 2002/0055694 A1 5/2002 Halperin et al.
 2002/0133197 A1 9/2002 Snyder et al.
 2002/0165471 A1 11/2002 Halperin et al.
 2002/0165585 A1 11/2002 Dupelle et al.
 2002/0193711 A1 12/2002 Halperin et al.
 2003/0055460 A1 3/2003 Owen et al.
 2003/0083699 A1 5/2003 Hamilton et al.
 2003/0195567 A1 10/2003 Jayne et al.
 2004/0044374 A1 3/2004 Weinberg et al.
 2004/0049234 A1 3/2004 Morgan et al.
 2004/0058305 A1 3/2004 Lurie et al.
 2004/0082888 A1 4/2004 Palazzolo et al.
 2004/0162585 A1 8/2004 Elghazzawi et al.
 2004/0210171 A1 10/2004 Palazzolo et al.
 2004/0215244 A1 10/2004 Marcovecchio et al.
 2005/0119706 A1 6/2005 Ideker et al.
 2005/0256415 A1 11/2005 Tan et al.
 2006/0036292 A1 2/2006 Smith et al.
 2006/0116724 A1 6/2006 Snyder
 2007/0032729 A1 2/2007 Fortin
 2007/0032829 A1 2/2007 Ostroff

FOREIGN PATENT DOCUMENTS

EP 1491176 12/2004
 GB 9713345 6/1997
 WO 97/24062 7/1997

(56)

References Cited

FOREIGN PATENT DOCUMENTS

WO	98/30282	7/1998
WO	99/24114	5/1999
WO	99/25306	5/1999
WO	01/56652	8/2001
WO	01/66182	9/2001
WO	02/15836	2/2002
WO	03/009895	2/2003
WO	WO 03/020364	3/2003
WO	2004/054656	7/2004
WO	2005/021089	3/2005

OTHER PUBLICATIONS

Aase et al., "CPR Artifact Removal from Human ECG Using Optimal Multichannel Filtering," IEEE Transactions on Biomedical Engineering, vol. 47, 1440-1449, (2000).

Afonso et al., "Detecting Ventricular Fibrillation", IEEE Engineering in Medicine and Biology, vol. 14:2, pp. 152-159 (1995).

Al-Fahoum et al., "Combined wavelet transformation and radial basis neural networks for classifying life-threatening cardiac arrhythmias", Medical & Biological Engineering & Computing, vol. 37:5, pp. 566-573 (1999).

Amann et al., Reliability of Fibrillation Detection Algorithms in Automatic External Defibrillators (AEDs), Dept. of Anaesthesia and Intensive Care Medicine, Leopold-Franzens-Universitat Innsbruck, Anichstr. 35, A-6020 Innsbruck, Austria, Dept. of Computer Science, Applied Mathematics Group, FH-Vorarlberg, Achstr. 1, A-6850 Dornbirn, Austria. At the top of the paper I have is the following: Jahrestagung der Österreichischen Deutschen und Schweizerischen Gesellschaft für Biomedizinische Technik Sep. 2003.

American Red Cross—Adult CPR/AED Training—Workplace Programs, <http://www.redcross.org/hss/cpraed.html>, printed from Internet May 14, 1999.

Barro et al., "Algorithmic sequential decision-making in the frequency domain for life threatening ventricular arrhythmias and imitative artifacts: a diagnostic system", J. Biomed. Eng., vol. 11:4, pp. 320-328 (1989).

Botsivaly et al., "Evaluation of a new technique for the Detection of Ventricular Fibrillation and Ventricular Tachycardia", Procs of the 22nd Ann EMBS Int Conf, Chicago, IL (2000).

Callaway et al., "Scaling exponent predicts defibrillation success for out-of-hospital ventricular fibrillation cardiac arrest," Circulation 103(12):1656-1661 (2001).

Callaway et al., "Ventricular Fibrillation Waveform Predicts Defibrillation Success by Automatic External Defibrillators", Academic Emergency Medicine, vol. 7:5, pp. 1-2 (2000).

Cardiac Science Brochure, Analysis Algorithm Overview, Powerheart® AED Automated External Defibrillator with RHYTHMx® Technology (no date).

Clayton et al., "Comparison of four techniques for recognition of ventricular fibrillation from the surface ECG", Medical & Biological Engineering & Computing, vol. 31:2, pp. 111-117 (1993).

Eftestol et al., "Effects of Interrupting Precordial Compressions on the Calculated Probability of Defibrillation Success During Out-of-Hospital Cardiac Arrest," Circulation, 105, 2270-2273, (2002).

Eftestol et al., "Predicting Outcome of Defibrillation by Spectral Characterization and Nonparametric Classification of Ventricular Fibrillation in Patients With Out-of-Hospital Cardiac Arrest", Circulation, 102:1523-1529 (2000).

Eftestol et al., "Probability of successful defibrillation as a monitor during CPR in out-of-hospital cardiac arrested patients," Resuscitation 48(3):245-254 (2001).

Fitzgibbon et al., "Determination of the noise source in the electrocardiogram during cardiopulmonary resuscitation", Crit Care Med, vol. 30:4, pp. S148-S152 (2002).

Flewelling, Nellcor Incorporated, Noninvasive Optical Monitoring, Chap. 88, pp. 1346-1353. CRC Press, Inc., 1995.

Force Sensing Resistors—An Overview of the Technology, FSR Integration Guide & Evaluation Parts Catalog with Suggested Electrical Interfaces (no date).

Ge et al., "Cardiac arrhythmia classification using autoregressive modeling", Biomed Eng. Online, pp. 13, (2002).

Geheb, Frederick J., "A System for the Determination of Ventricular Tachycardia or Ventricular Fibrillation during Cardio-Pulmonary Resuscitation", 2 pages (Apr. 2002).

Gruben et al., "System for Mechanical Measurements During Cardiopulmonary Resuscitation in Humans," IEEE Transactions on Biomedical Engineering, vol. 37, No. 2, Feb. 1990.

Haykin, Adaptive Filter Theory, Third Edition, Upper Saddle River, NJ, USA. Prentice-Hall, 1996.

Heartstream—The Background Behind Our Technology, <http://www.heartstream.com/techbk.htm>, printed from Internet Jun. 25, 1999.

Husoy et al., "Removal of Cardiopulmonary Resuscitation Artifacts from Human ECG Using an Efficient Matching Pursuit-Like Algorithm," IEEE Transactions on Biomedical Engineering, vol. 49, 1287-1298, (2002).

Khadra et al., "Detection of life-threatening cardiac arrhythmias using the wavelet transformation", Medical & Biological Engineering & Computing, vol. 35:5, pp. 626-632 (1997).

Kuo et al., "Computer Detection of Ventricular Fibrillation", Computers in Cardiology, pp. 347-349 (Sep. 1978).

Langhelle et al. "Reducing CPR Artifacts in Ventricular Fibrillation in Vitro," Resuscitation, Mar; 48(3):279-91 (2001).

Lightfoot et al., "Dynamic nature of electrocardiographic waveform predicts rescue shock outcome in porcine ventricular fibrillation," Ann. Emerg. Med. 42(2):230-41 (Aug. 2003).

Menegazzi et al., "Immediate defibrillation versus interventions first in a swine model of prolonged ventricular fibrillation", Resuscitation, vol. 59, pp. 261-270 (2003).

Menegazzi et al., "Ventricular Fibrillation Scaling Exponent Can Guide Timing of Defibrillation and Other Therapies", Circulation, 109:926-931 (Feb. 2004).

Noc et al., "Electrocardiograph Prediction of the Success of Cardiac Resuscitation," Critical Care Medicine, Williams and Wilkins Company, Baltimore, MA, US, vol. 27, No. 4, pp. 708-714 (Apr. 1, 1999).

Nygards et al., "Recognition of Ventricular Fibrillation Utilizing the Power Spectrum of the ECG", Computers in Cardiology, pp. 393-397 (1997).

Patterson, "Impedance cardiography: What is the source of the signal?", International Conf. on Electrical Bioimpedance, Journal of Physics: Conference Series 224 (2010).

Povoas et al., "Predicting the success of defibrillation by electrocardiographic analysis," Resuscitation 53(1):77-82 (2002).

Sherman et al., "Ventricular fibrillation exhibits dynamical properties and self-similarity", Resuscitation, vol. 47, pp. 163-173 (2000).

Strohmenger, "Kammerflimmern/-flattern: Prädiktoren für den Defibrillationserfolg?" Notfall & Rettungsmedizin, vol. 6, pp. 22-26 (Feb. 1, 2003) (English abstract).

U.S. Appl. No. 10/421,652 (Marcovecchio, Optical Pulse Sensor for External Defibrillator).

U.S. Appl. No. 10/370,036 (Elghazzawi et al., CPR Sensitive ECG Analysis in an Automatic External Defibrillator).

U.S. Appl. No. 10/441,933 (Marcovecchio, Processing Pulse Signal in Conjunction with ECG Signal).

Wang et al., "Effects of Biphasic vs Monophasic Defibrillation on the Scaling Exponent in a Swine Model of Prolonged Ventricular Fibrillation", Academic Emergency Medicine, vol. 8:8, pp. 771-780 (2001).

Watson et al., "A novel wavelet transform based analysis reveals hidden structure in ventricular fibrillation", Resuscitation, vol. 43:2, pp. 121-127 (2000).

Yoji et al., "Adverse effects of interrupting precordial compression during cardiopulmonary resuscitation", Critical Care Medicine, vol. 25:5, pp. 733-736 (1997).

Yu et al., "Adverse Outcomes of Interrupted Precordial Compression During Automated Defibrillation," Circulation, 106, 368-372 (2002).

(56)

References Cited

OTHER PUBLICATIONS

Pinchak et al., "Chest wall acceleration and force measurements in simulated manual and mechanical cardiopulmonary resuscitation," *Critical Care Medicine*, vol. 16, No. 2, pp. 151-160 (1988).
Johnston et al., "The Potential Use of Impedance Cardiography as a Hemodynamic Sensor for Automated External Defibrillators," *Abstracts* 959-104 (Feb. 1995).

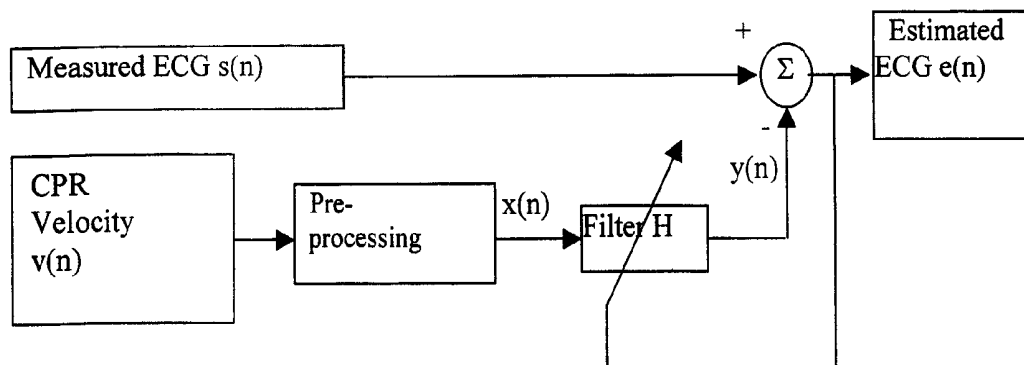


FIG. 1

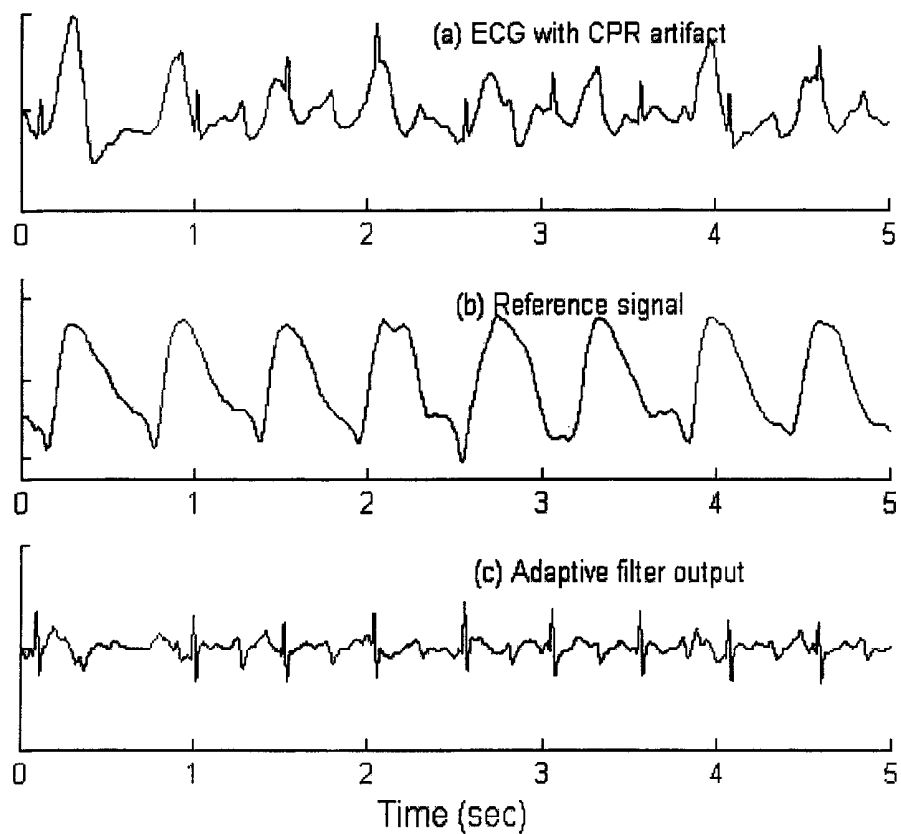


FIG. 2

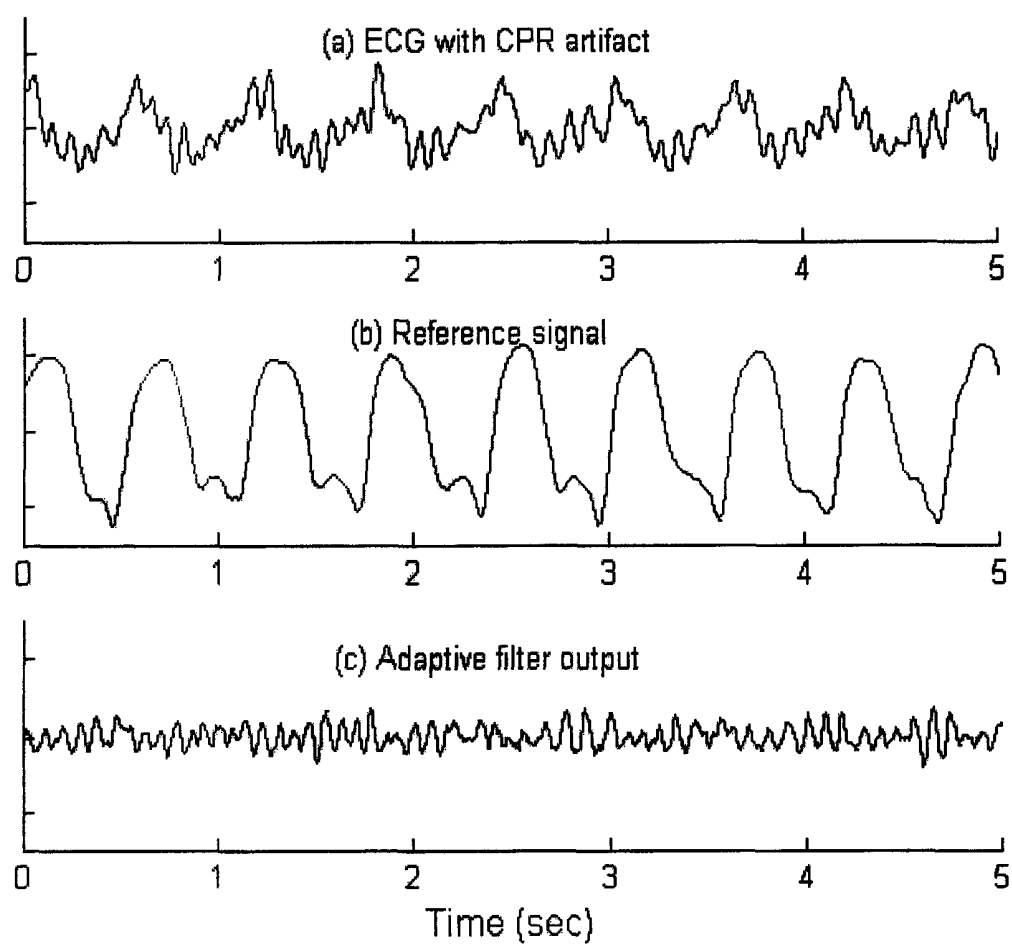


FIG. 3

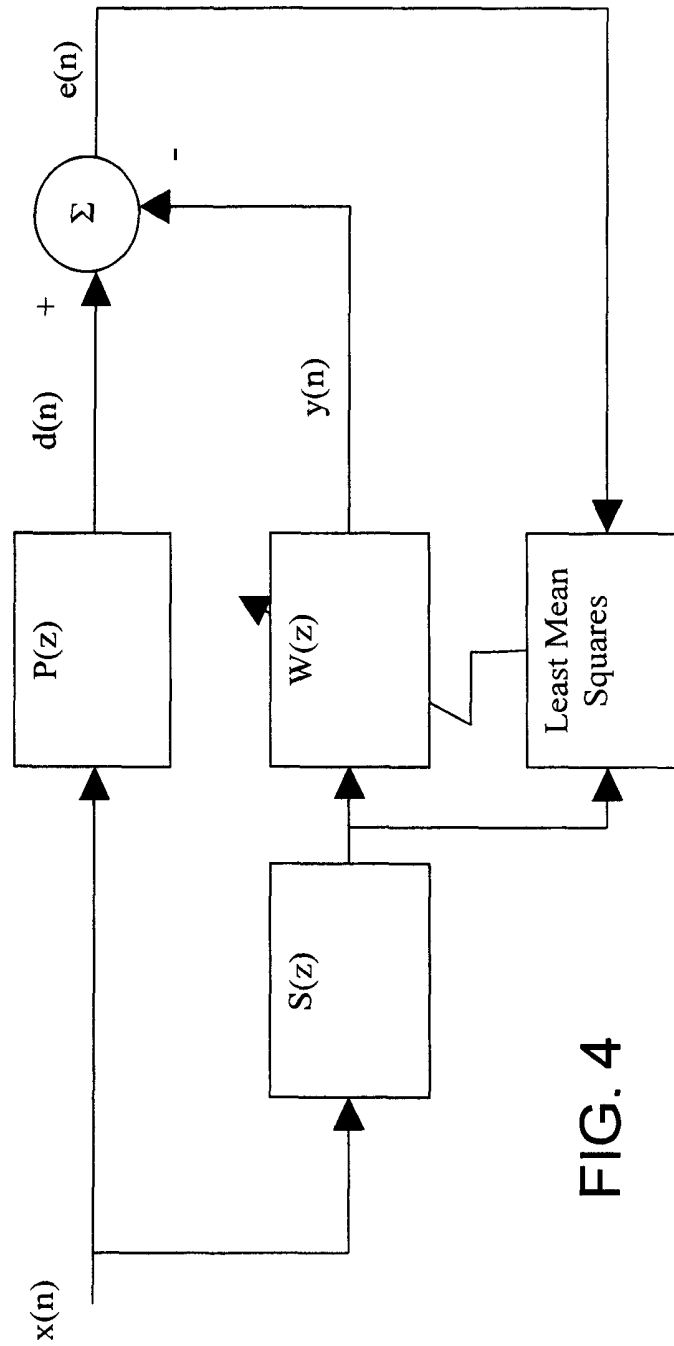


FIG. 4

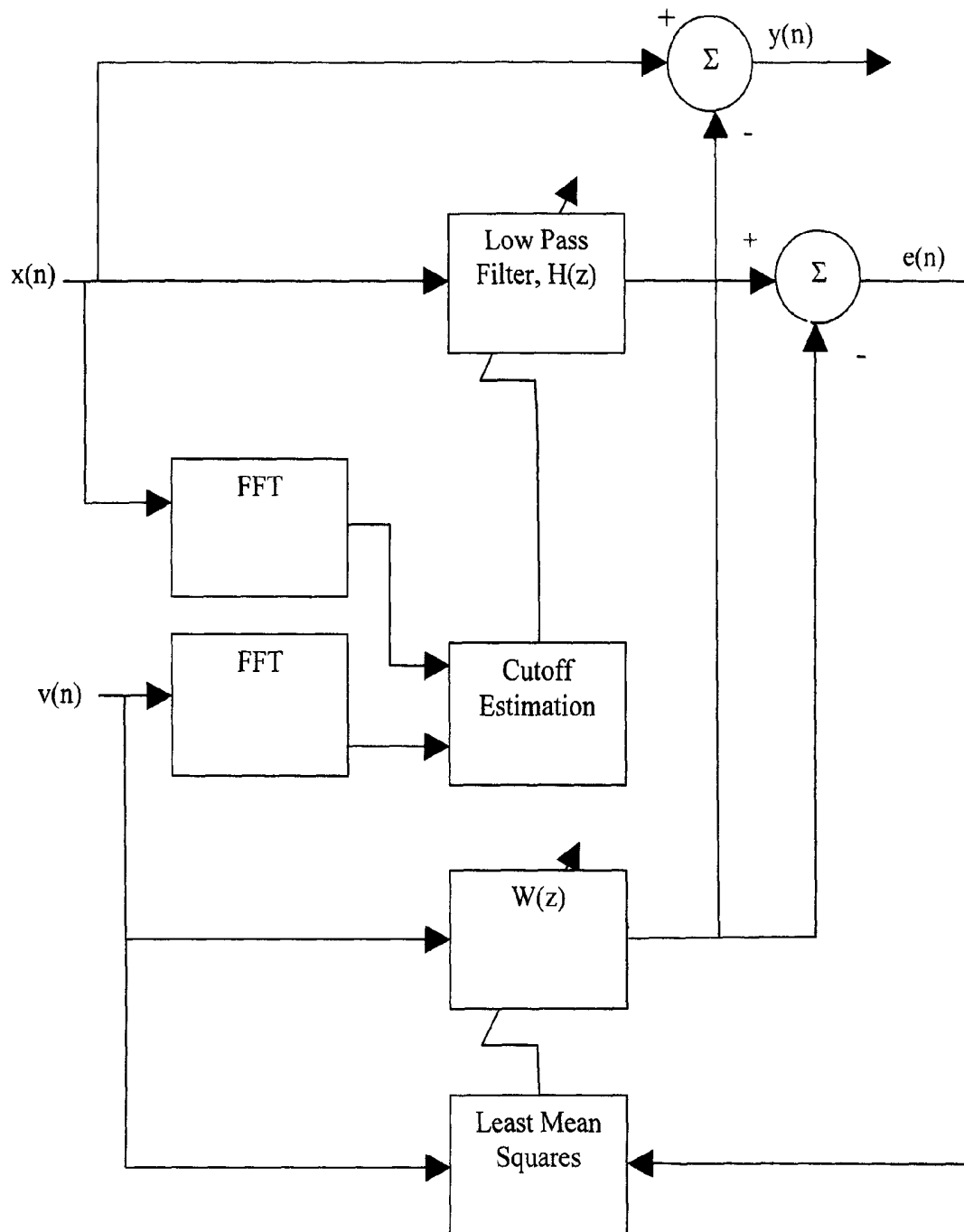


FIG. 5

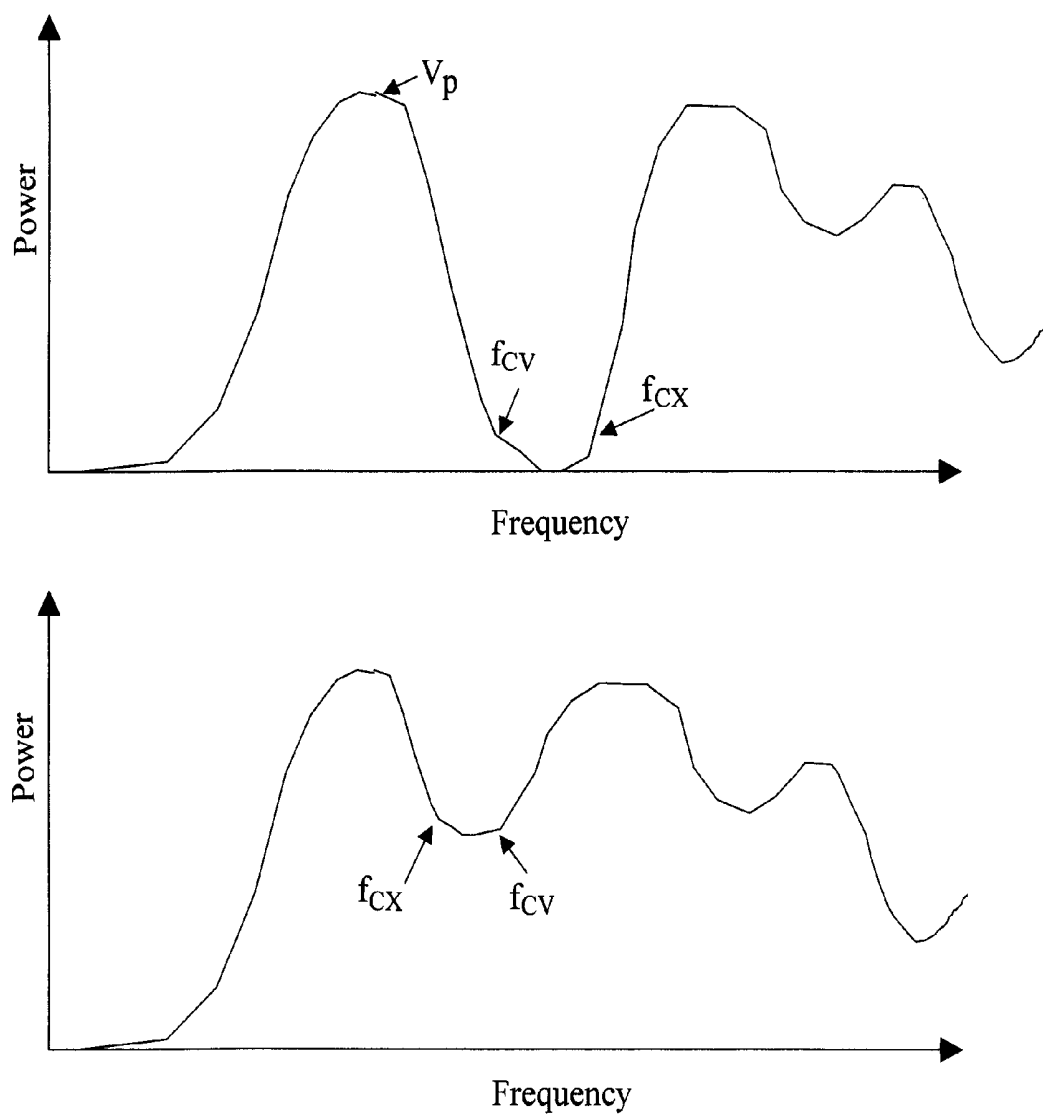


FIG. 6

USING CHEST VELOCITY TO PROCESS PHYSIOLOGICAL SIGNALS TO REMOVE CHEST COMPRESSION ARTIFACTS

CROSS REFERENCE TO RELATED APPLICATION

This application is a continuation application of and claims priority to U.S. application Ser. No. 13/872,033, filed Apr. 26, 2013, which application is a continuation application of and claims priority to U.S. application Ser. No. 10/786,359, filed Feb. 24, 2004, now abandoned, which application is a continuation-in-part of and claims priority to U.S. application Ser. No. 10/704,366, filed on Nov. 6, 2003, now issued U.S. Pat. No. 7,220,235. All are hereby incorporated by reference.

TECHNICAL FIELD

This invention relates to devices for assisting cardiac resuscitation.

BACKGROUND

Resuscitation treatments for patients suffering from cardiac arrest generally include clearing and opening the patient's airway, providing rescue breathing for the patient, and applying chest compressions to provide blood flow to the victim's heart, brain and other vital organs. If the patient has a shockable heart rhythm, resuscitation also may include defibrillation therapy. The term basic life support (BLS) involves all the following elements: initial assessment; airway maintenance; expired air ventilation (rescue breathing); and chest compression. When all three (airway breathing, and circulation, including chest compressions) are combined, the term cardiopulmonary resuscitation (CPR) is used.

Current automated ECG rhythm analysis methods interrupt cardiopulmonary resuscitation (CPR) to avoid artifacts in the ECG resulting from chest compressions. Long interruptions of CPR have been shown to result in higher failure rate of resuscitation. Studies have reported that the discontinuation of precordial compression can significantly reduce the recovery rate of spontaneous circulation and the 24-hour survival rate. Y. Sato, M H. Weil, S. Sun, W. Tang, J. Xie, M. Noc, and J. Bisera, Adverse effects of interrupting precordial compression during cardiopulmonary resuscitation, *Critical Care Medicine*, Vol. 25(5), 733-736 (1997). Yu et al., 2002. *Circulation*, 106, 368-372 (2002), T. Eftestol, K. Sunde, and P A. Steen, Effects of Interrupting Precordial Compressions on the Calculated Probability of Defibrillation Success During Out-of-Hospital Cardiac Arrest, *Circulation*, 105, 2270-2273, (2002).

Management of breathing is another important aspect of the CPR process. Typical methods of monitoring breathing employ some form of impedance pneumography which measure and track changes in the transthoracic impedance of the patient. Currently, however, chest compressions result in significant artifact on the impedance signals, resulting in impedance-based pneumographic techniques as unreliable indicators of lung volume during chest compressions.

Adaptive filters have been attempted as a way of removing chest-compression artifacts in the ECG signal. S O. Aase, T. Eftestol, J H. Husoy, K. Sunde, and P A. Steen, CPR Artifact Removal from Human ECG Using Optimal Multi-channel Filtering, *IEEE Transactions on Biomedical Engineering*, Vol. 47, 1440-1449, (2000). A. Langhelle, T. Eft-

estol, H. Myklebust, M. Eriksen, B T. Holten, P A. Steen, Reducing CPR Artifacts in Ventricular Fibrillation in Vitro. *Resuscitation*. March; 48(3):279-91 (2001). J H. Husoy, J. Eilevstjonn, T. Eftestol, S O. Aase, H Myklebust, and P A. Steen, Removal of Cardiopulmonary Resuscitation Artifacts from Human ECG Using an Efficient Matching Pursuit-Like Algorithm, *IEEE Transactions on Biomedical Engineering*, Vol 49, 1287-1298, (2002). H R. Halperin, and R D. Berger, CPR Chest Compression Monitor, U.S. Pat. No. 6,390,996 (2002). Aase et al. (2000) and Langhelle et al. (2001) used the compression depth and thorax impedance as reference signals for their adaptive filter. Husoy et al. (2002) extended this study by using a matching pursuit iteration to reduce the computational complexity; however, their results are usually computationally intensive, such as involving the calculation of a high order inverse filter. Halperin et al. (2002) proposed a frequency-domain approach using the auto- and the cross-spectrum of the signals and a time-domain approach using a recursive least square method for adaptive filtering the ECG signal. In both approaches, intensive computations are required.

There are numerous references available on adaptive filters. E.g., S. Haykin, *Adaptive Filter Theory*, Third Edition, Upper Saddle River, N.J., USA. Prentice-Hall, 1996

SUMMARY

In general the invention features a method of analyzing a physiological (e.g., an ECG) signal during application of chest compressions. The method includes acquiring a physiological signal during application of chest compressions; acquiring the output of a sensor from which information on the velocity of chest compressions can be determined; and using the information on the velocity to reduce at least one signal artifact in the physiological signal resulting from the chest compressions.

Preferred implementations of the invention may incorporate one or more of the following: The physiological signal may be any of a variety of physiological signals, including an ECG signal, an IPG signal, an ICG signal, or a pulse oximetry signal. The sensor may be a velocity sensor, and the information on the velocity may be determined from the velocity sensor. The sensor may be an accelerometer, and the information on the velocity may be determined from integration of the output of the accelerometer. Using the information on the velocity to reduce at least one signal artifact in the physiological signal may comprise time aligning the physiological signal with the velocity. Using the information on the velocity to reduce at least one signal artifact in the physiological signal may comprise using an adaptive filter that may be adjusted to remove chest compression artifacts. The method may include a ventricular fibrillation detection algorithm for processing the physiological signal with reduced artifact to estimate whether a ventricular fibrillation may be present. The method may include a preprocessing step that detects when chest compressions are applied and automatically initiates the adaptive filter. The method may include enabling delivery of a defibrillation shock if the algorithm estimates that ventricular fibrillation is present. A difference signal may be produced, the difference signal being representative of the difference between the physiological signal fed into the adaptive filter and the physiological signal after artifact reduction by the adaptive filter. The difference signal may provide a measure of the amount of artifact in the physiological signal. The difference signal may be used to modify the subsequent processing of the physiological signal. If the difference signal indicates that

the amount of artifact exceeds a first threshold, the ventricular fibrillation detection algorithm may be modified to make it more resistant to being influenced by the artifact. If the difference signal indicates that the amount of artifact exceeds a second threshold higher than the first threshold, use of the ventricular defibrillation detection algorithm may be suspended. Spectral analysis may be performed on the difference signal, and adjustments may be made to filtering of the physiological signal based on the outcome of the spectral analysis. The velocity signal may undergo a normalization pre-processing prior to being fed to an adaptive filter. The adaptive filter may include an FIR filter. The adaptive filter may include a zero-th order filter. The adaptive filter may have coefficients that are dynamically controlled by an estimate of the physiological signal. The adaptive filter may have the capability of being automatically reset when the difference between the filter output and the measured physiological signal is beyond a threshold. The automatic reset may be capable of dynamically changing the step size and thus improving the relationship of convergence and stability of the filter. A time-aligning process may be performed on the physiological and velocity signals, wherein the time aligning process aligns the two signals relative to the compressions. The method may include adaptive filtering of the output of the time aligning process, wherein the adaptive filtering reduces the error between the physiological and velocity signals. The adaptive filter may include a Kalman filter. The adaptive filter may employ adaptive equalization.

Among the many advantages of the invention (some of which may be achieved only in some of its various implementations) are the following:

This invention provides excellent techniques for (a) adaptively removing the artifacts induced by CPR in an ECG signal, (b) enhancing an ECG signal for monitoring, and (c) increasing the reliability of ECG rhythm advisory algorithms.

As part of a rhythm advisory algorithm, various implementations of the invention could be incorporated in an ECG monitor, an external defibrillator, an ECG rhythm classifier, or a ventricular arrhythmia detector.

The invention makes it possible to continue performing CPR while ECG data is collected for an ECG rhythm advisory algorithm. This can enhance the result of CPR, leading, for example, to an increase in the success rate of resuscitation.

The invention can also provide a "cleansed" ECG signal output for display to the user of a defibrillator.

The invention also provides for the first time a means of measuring lung volume during chest compressions by impedance-based methods. The method may also be used to filter other physiological signals corrupted by compression-induced artifact, such as impedance cardiography and pulse oximetry.

This invention demonstrates excellent performance at removing the CPR artifact with a zero-th order FIR filter, thus making some implementations of the invention much simpler and faster than the adaptive-filter structures proposed in the prior art.

Pre-processing of the reference signal and an automatic-reset feature make it possible for some implementations of the invention to use a relatively large step size for adaptation, thus making convergence faster and more stable.

Some implementations of the invention achieve excellent performance in CPR-artifact removal at reduced computational cost.

Other features and advantages of the invention are described in the detailed description, drawings, and claims.

BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1 is a block diagram of one implementation of the invention.

FIG. 2 shows plots of the ECG signal, CPR reference signal, and output of adaptive filter for a normal sinus rhythm.

FIG. 3 shows plots of the ECG signal, CPR reference signal, and output of adaptive filter for ventricular fibrillation.

FIG. 4 is a block diagram of a filtered-X least mean squares (FXLMS ANC) algorithm.

FIG. 5 is a block diagram of an implementation using the algorithm of FIG. 4.

FIG. 6 shows two spectral power distributions related to the implementation of FIG. 5.

DETAILED DESCRIPTION

There are a great many possible implementations of the invention, too many to describe herein. Some possible implementations that are presently preferred are described below. It cannot be emphasized too strongly, however, that these are descriptions of implementations of the invention, and not descriptions of the invention, which is not limited to the detailed implementations described in this section but is described in broader terms in the claims.

One possible implementation is illustrated by a flow chart in FIG. 1. The front end of an AED acquires both the ECG signal and the CPR signal, which is the velocity of compression of the chest. If chest displacement or acceleration are measured instead of velocity, velocity can be mathematically acquired via one or more integration or differentiation operations from the measurement signal.

The velocity signal undergoes pre-processing, and is then fed to an adaptive filter. In a preferred implementation, the pre-processing is a normalization of the velocity signal so that the signal supplied to the adaptive filter is limited to be within 0 and 1. But normalization is not required. In another implementation, a time-aligning process is performed on the ECG and the reference signal by such methods as cross-correlation. This provides alignment of the two signals relative to the compressions so that the input signals of the adaptive filter are better aligned. But this aligning process is not required. Other preprocessing can be applied to the velocity signal to improve the performance of the adaptive filter.

In FIG. 1, $x(n)$ and $y(n)$ are the input and the output of the adaptive filter H , which can be an FIR filter, an IIR filter, or another type of filter. In a preferred implementation, the coefficients of the filter are dynamically controlled by the estimated ECG signal:

$$h(n) = h(n-1) + m \times e(n) \times X(n)$$

where $h(n)$ is a vector containing the filter coefficients, m is a vector containing the step sizes for each filter coefficients, $e(n)$ is the estimated ECG signal, and $X(n)$ is a vector containing the input data. The estimated ECG signal is computed by subtracting the filter output $y(n)$ from the measured ECG signal (containing artifact).

In some implementations, there is an automated resetting mechanism. When the difference between the filter output

$y(n)$ and the measured ECG $s(n)$ is beyond a threshold, the adaptive filter will reset its coefficients so that the system will not become unstable.

Other filter structures than the one shown in FIG. 1, as well as other mathematical representations of the filtering, are possible.

FIG. 2 shows samples of the performance of the adaptive filter of FIG. 1 in response to a normal sinus rhythm. The signal in (a) is the ECG signal with CPR artifact. The signal in (b) is the compression velocity used as the reference signal. The signal in (c) is the output of the adaptive filter.

FIG. 3 shows samples of the performance of the adaptive filter of FIG. 1 during ventricular fibrillation. The signal in (a) is the ECG signal with CPR artifact. The signal in (b) is the compression velocity used as the reference signal. The signal in (c) is the output of the adaptive filter.

As shown in both FIG. 2 and FIG. 3, the implementation of FIG. 1 is able to suppress the CPR artifacts embedded in the measured ECG signals (a). The CPR artifact is nearly, if not completely, removed in the estimated ECG signal (c). The velocity signal (b) used as a reference signal is clearly correlated with the CPR artifacts in the measured ECG signals (a).

The adaptive filter assumes that the artifact in the signal is correlated with the reference signal and uncorrelated with the desired signal (estimated ECG). It thus adaptively estimates the artifact using the reference signal and subtracts the estimated artifact from the measured ECG signal.

The results shown in FIG. 2 are based on a 0th-order FIR filter, which simply scales the current sample of the ECG signal adaptively. The CPR artifact was significantly reduced, if not completely removed. This implementation thus combines simplicity and efficiency in its performance.

In the applications of adaptive filters, the speed of adaptation convergence is usually controlled by a step-size variable. A faster convergence requires a larger step size, which usually tends to make the filter less stable. The automatic resetting mechanism of some implementations can dynamically change the step size and thus improve the relation of convergence and stability.

The coefficients of the filter are updated in a sample-by-sample manner. The changes of the coefficients, i.e., $h(n) - h(n-1)$ is proportional to the product of the step size and the reference signal. The amplitude of the reference signal can thus affect the stability and convergence of the filter. The pre-processing of the reference signal can therefore enhance the performance of the filter by adjusting the reference signal.

In another implementation, a time-aligning process is performed on the ECG and velocity signals by such methods as cross-correlation. This provides alignment of the two signals relative to the compressions. Then, preferably, adaptive filtering methods are used such as those involved in the minimization of the mean-squared error between the ECG and the velocity.

A processing unit could be provided for detecting when compressions are being applied and automatically turning on the adaptive filter. The output of the adaptive filter (i.e., the ECG signal with artifact reduced) could be supplied to a ventricular fibrillation (VF) detection algorithm (e.g., a shock advisory algorithm) of an automatic external defibrillator (AED).

An error signal could be produced that is representative of the difference between the ECG input and ECG output of the adaptive filter. This error signal would give a measure of the amount of CPR artifact in the signal, and it would be useful as a means of modifying the subsequent processing of the

ECG. For instance, if the artifact level gets high enough (e.g., higher than a first threshold), the VF detection algorithm thresholds could be increased to make it more resistant to any CPR artifact that still remained in the ECG signal. If the level got even higher (e.g., higher than a second threshold higher than the first threshold), the VF detection could be shut off entirely.

In preferred implementation, the filter output is presented graphically on the display of a defibrillator or other medical device incorporating an electro-cardiographic function. The filter output may also be printed on a strip-chart recorder in the medical device. Alternatively, the filter output may provide the input signal for subsequent signal processing performed by the processing means. The purpose of such signal processing may take the form of QRS detection, paced beat detection during pacing, arrhythmia analysis, and detection of ventricular fibrillation or other shockable rhythms.

Spectral analysis could be performed on the error signal, and based on the major bands of frequency content of the error signal, the pre-filtering of the ECG signal prior to the VF detection can be adjusted. For instance, if the error signal is found to reside primarily in the 3-5 Hz band, additional filtering can be provided in that band prior to input into the VF detection (or other ECG processing) algorithm.

Many other implementations of the invention other than those described above are within the invention, which is defined by the following claims.

For example, methods of adaptive channel equalization may be employed to ameliorate both synchronization and phase errors in the velocity waveform. Kalman filtering techniques may also be employed to improve performance of the filter when rescuer performance of chest compressions changes over time and is better modeled as a non-stationary process.

Time alignment of the ECG and velocity signal may also be accomplished by such methods as cross-correlation techniques known to those skilled in the art. This will provide alignment of the two signals relative to the compressions. Then, preferably, adaptive filtering methods are used such as those involved in the minimization of the mean-squared error between the ECG and the velocity.

In a further implementation, more sophisticated signal processing methods may be used to minimize ECG artifacts induced by CPR chest compressions. For example, methods known as feed forward active noise cancellation (FANC) may be used. FIG. 4 shows a block diagram of the filtered-X least mean squares (FXLMS ANC) algorithm, as developed by Widrow and Burgess. $P(z)$ represents the unknown plant through which the signal $x(n)$ is filtered. Digital filter $W(z)$ is adaptively adjusted to minimize the error signal $e(n)$. In one implementation, as depicted in FIG. 5, $x(n)$ is the unfiltered ECG signal, $P(z)$ is eliminated from the diagram, and $d(n)$ is approximated with the chest compression velocity signal $v(n)$. In the LMS algorithm, assuming a mean square cost function $\zeta(n) = E[e^2(n)]$, the adaptive filter minimizes the instantaneous squared error, $\zeta(n) = e^2(n)$, using the steepest descent algorithm, which updates the coefficient vector in the negative gradient direction with step size μ :

$$w(n+1) = w(n) - \mu/2 * \tilde{\nabla} \zeta(n),$$

where $\tilde{\nabla} \zeta(n)$ is an instantaneous estimate of the mean square error (MSE) gradient at time n equal to $-2v(n)e(n)$. Stability and accuracy of the FXLMS ANC algorithm can be improved by adding a variable cutoff low pass filter $H(z)$ to eliminate frequency components in the ECG not related to the chest compression artifact. In general, the spectral

energy of the chest compression artifact is predominately lower than those of the ECG. A cutoff frequency of approximately 3 Hz is adequate in many cases, but this may vary from patient to patient and among different rescuers performing chest compressions. To overcome this difficulty, an FFT is performed on $v(n)$ and input into a cutoff frequency estimation (CFE) procedure that determines the optimal cutoff frequency, f_C , for the lowpass filter. In a preferred implementation, the decision is based on calculating the frequency, not to exceed 5 Hz, below which 80% of the waveform energy is present, but this percentage may vary and additional decision logic may be employed. For instance, an FFT may also be calculated for $x(n)$, also input to the CFE procedure. By first normalizing amplitude of the frequency spectra $X(z)$ amplitude peak of the compression artifact and then subtracting the velocity spectra $V(z)$ from the normalized input $X'(z)$, the difference spectra is calculated $\Delta X'(z) = X'(z) - V'(z)$. Frequencies are then determined for $V(z)$ and $\Delta X'(z)$ at which most of the spectral energy is within, set in this embodiment to 97%, and labeled f_{CV} and f_{CX} , respectively, and shown in FIG. 6. f_C is then set to the lesser of f_{CV} and f_{CX} . Alternatively, f_C can be set to some intermediate frequency between f_{CV} and f_{CX} .

The quality of other physiological signals, such as impedance cardiographic (ICG), impedance pneumographic (IPG), or pulse oximetry, known to those skilled in the art, may also be also be enhanced by the filter, particularly if the sensor is located on the thoracic cage in nearby proximity to the motion sensor from which the velocity signal is derived. Minimization of compression artifact with impedance pneumography signals can be accomplished with any of the previously described methods.

The adaptive filter can be used to minimize the cross-correlation of the adaptive-filter output with the reference signal or the cross-correlation of the adaptive-filter output with the measured ECG signal.

The invention claimed is:

1. A method of analyzing a physiological signal during application of chest compressions, the method comprising: acquiring a physiological signal during application of chest compressions; acquiring the output of a sensor from which information on the velocity of chest compressions can be determined; using the information on the velocity to reduce at least one signal artifact in the physiological signal resulting from the chest compressions; and using a ventricular fibrillation detection algorithm for processing the physiological signal with reduced artifact to estimate whether a ventricular fibrillation is present, wherein using the information on the velocity to reduce at least one signal artifact in the physiological signal comprises using an adaptive filter that is adjusted to remove chest compression artifacts, wherein a difference signal is produced, the difference signal being representative of the difference between the physiological signal fed into the adaptive filter and the physiological signal after artifact reduction by the adaptive filter, wherein the difference signal provides a measure of the amount of artifact in the physiological signal, and further comprising the step of using the difference signal to modify the subsequent processing of the physiological signal, wherein, if the difference signal indicates that the amount of artifact exceeds a first threshold, the ventricular

fibrillation detection algorithm is modified to make it more resistant to being influenced by the artifact.

2. The method of claim 1 wherein, if the difference signal indicates that the amount of artifact exceeds a second threshold higher than the first threshold, use of the ventricular fibrillation detection algorithm is suspended.

3. A method of analyzing a physiological signal during application of chest compressions, the method comprising: acquiring a physiological signal during application of chest compressions;

acquiring the output of a sensor from which information on the velocity of chest compressions can be determined; and

using the information on the velocity to reduce at least one signal artifact in the physiological signal resulting from the chest compressions,

wherein using the information on the velocity to reduce at least one signal artifact in the physiological signal comprises using an adaptive filter that is adjusted to remove chest compression artifacts,

wherein a difference signal is produced, the difference signal being representative of the difference between the physiological signal fed into the adaptive filter and the physiological signal after artifact reduction by the adaptive filter, and

wherein the adaptive filter comprises the capability of being automatically reset when the difference signal exceeds a threshold.

4. A method of analyzing a physiological signal during application of chest compressions, the method comprising: acquiring a physiological signal during application of chest compressions;

acquiring the output of a sensor from which information on the velocity of chest compressions can be determined; and

using the information on the velocity to reduce at least one signal artifact in the physiological signal resulting from the chest compressions,

wherein using the information on the velocity to reduce at least one signal artifact in the physiological signal comprises using an adaptive filter that is adjusted to remove chest compression artifacts, and

wherein the adaptive filter comprises the capability of dynamically changing a step size and thus improving a relationship of convergence and stability of the filter.

5. A method of analyzing a physiological signal during application of chest compressions, the method comprising: acquiring a physiological signal during application of chest compressions;

acquiring the output of a sensor from which information on the velocity of chest compressions can be determined; and

using the information on the velocity to reduce at least one signal artifact in the physiological signal resulting from the chest compressions,

wherein using the information on the velocity to reduce at least one signal artifact in the physiological signal comprises using an adaptive filter that is adjusted to remove chest compression artifacts,

further comprising a ventricular fibrillation detection algorithm for processing the physiological signal with reduced artifact to estimate whether a ventricular fibrillation is present,

further comprising a preprocessing step that detects when chest compressions are applied and automatically initiates the adaptive filter,

further comprising a time-aligning process performed on the physiological and velocity signals, wherein the time aligning process aligns the two signals relative to the compressions, and

further comprising adaptive filtering of the output of the time aligning process, wherein the adaptive filtering reduces the error between the physiological and velocity signals.

6. A method of analyzing a physiological signal during application of chest compressions, the method comprising: acquiring a physiological signal during application of chest compressions; acquiring the output of a sensor from which information on the velocity of chest compressions can be determined; and using the information on the velocity to reduce at least one signal artifact in the physiological signal resulting from the chest compressions, wherein using the information on the velocity to reduce at least one signal artifact in the physiological signal comprises using an adaptive filter that is adjusted to remove chest compression artifacts and wherein the adaptive filter employs adaptive equalization.

7. The method of claim 6 further comprising a ventricular fibrillation detection algorithm for processing the physiological signal with reduced artifact to estimate whether a ventricular fibrillation is present.

8. The method of claim 7 further comprising a pre-processing step that detects when chest compressions are applied and automatically initiates the adaptive filter.

9. The method of claim 8 wherein the physiological signal is an ICG signal.

10. The method of claim 8 wherein the physiological signal is a pulse oximetry signal.

11. The method of claim 8 wherein the sensor is a velocity sensor, and the information on the velocity is determined from the velocity sensor.

12. The method of claim 8 wherein the sensor is an accelerometer, and the information on the velocity is determined from integration of the output of the accelerometer.

13. The method of claim 8 wherein using the information on the velocity to reduce at least one signal artifact in the physiological signal comprises time aligning the physiological signal with the velocity.

14. The method of claim 8 further comprising enabling delivery of a defibrillation shock if the algorithm estimates that ventricular fibrillation is present.

15. The method of claim 8 further comprising a time-aligning process performed on the physiological and velocity signals, wherein the time aligning process aligns the two signals relative to the compressions.

16. The method of claim 15 further comprising adaptive filtering of the output of the time aligning process, wherein the adaptive filtering reduces the error between the physiological and velocity signals.

17. The method of claim 8 wherein the physiological signal is an ECG signal.

18. The method of claim 8 wherein the physiological signal is an IPG signal.

19. The method of claim 6 wherein a difference signal is produced, the difference signal being representative of the difference between the physiological signal fed into the adaptive filter and the physiological signal after artifact reduction by the adaptive filter.

20. The method of claim 19 wherein the difference signal provides a measure of the amount of artifact in the physiological signal.

21. The method of claim 20 further comprising the step of using the difference signal to modify the subsequent processing of the physiological signal.

22. The method of claim 21 wherein, if the difference signal indicates that the amount of artifact exceeds a first threshold, the ventricular fibrillation detection algorithm is modified to make it more resistant to being influenced by the artifact.

23. The method of claim 22 wherein, if the difference signal indicates that the amount of artifact exceeds a second threshold higher than the first threshold, use of the ventricular defibrillation detection algorithm is suspended.

24. The method of claim 21 wherein spectral analysis is performed on the difference signal, and adjustments are made to filtering of the physiological signal based on the outcome of the spectral analysis.

25. The method of claim 6 wherein the velocity signal undergoes a normalization pre-processing prior to being fed to an adaptive filter.

26. The method of claim 6 wherein the adaptive filter comprises an FIR filter.

27. The method of claim 26 wherein the adaptive filter comprises a zero-th order filter.

28. The method of claim 6 wherein the adaptive filter comprises coefficients that are dynamically controlled by an estimate of the physiological signal.

29. The method of claim 6 wherein the adaptive filter comprises the capability of being automatically reset when the difference between the filter output and the measured physiological signal is beyond a threshold.

30. The method of claim 29 wherein the automatic reset comprises the capability of dynamically changing the step size and thus improving the relationship of convergence and stability of the filter.

31. The method of claim 6 wherein the adaptive filter comprises a Kalman filter.

* * * * *

专利名称(译)	使用胸部速度处理生理信号以消除胸部压缩伪影		
公开(公告)号	US9521978	公开(公告)日	2016-12-20
申请号	US14/512167	申请日	2014-10-10
[标]申请(专利权)人(译)	卓尔医学产品公司		
申请(专利权)人(译)	ZOLL医疗公司		
当前申请(专利权)人(译)	ZOLL医疗公司		
[标]发明人	FREEMAN GARY A TAN QING GEHEB FREDERICK J		
发明人	FREEMAN, GARY A. TAN, QING GEHEB, FREDERICK J.		
IPC分类号	A61N1/39 A61H31/00 A61B5/046 A61B5/00 A61B5/04 A61B5/0205 A61B5/1455 A61B5/11 A61B5/0402		
CPC分类号	A61B5/721 A61B5/0205 A61B5/046 A61B5/04017 A61B5/11 A61B5/14551 A61B5/4836 A61B5/4848 A61B5/7207 A61H31/005 A61H31/006 A61H31/007 A61N1/3987 A61H2201/5058 A61H2201/5079 A61H2201/5084 A61H2230/04 A61N1/3925 A61B5/0402 A61H2201/5043 A61H2201/5048 A61N1/046 A61N1/0492 A61N1/3993		
代理机构(译)	FISH & RICHARDSON P.C.		
其他公开文献	US20150031972A1		
外部链接	Espacenet USPTO		

摘要(译)

一种在应用胸部按压期间分析生理 (例如, ECG) 信号的方法。该方法包括: 在施加胸部按压期间获取生理信号; 获取传感器的输出, 从该传感器可以确定关于胸部按压的速度的信息; 以及使用关于所述速度的信息以减少由所述胸部按压产生的所述生理信号中的至少一个信号伪影。

