

University of Missouri, St. Louis

IRL @ UMSL

---

UMSL Patents

---

2-28-2017

## Method and system for non-invasive optical blood glucose detection utilizing spectral data analysis

Zhi Xu

zhixu@umsl.edu

Follow this and additional works at: <https://irl.umsl.edu/patents>

---

### Recommended Citation

Xu, Zhi, "Method and system for non-invasive optical blood glucose detection utilizing spectral data analysis" (2017). *UMSL Patents*. 48.  
<https://irl.umsl.edu/patents/48>

This Book is brought to you for free and open access by IRL @ UMSL. It has been accepted for inclusion in UMSL Patents by an authorized administrator of IRL @ UMSL. For more information, please contact [marvinh@umsl.edu](mailto:marvinh@umsl.edu).



(12) **United States Patent**  
**Xu**

(10) **Patent No.:** **US 9,579,049 B2**  
(45) **Date of Patent:** **Feb. 28, 2017**

(54) **METHOD AND SYSTEM FOR  
NON-INVASIVE OPTICAL BLOOD GLUCOSE  
DETECTION UTILIZING SPECTRAL DATA  
ANALYSIS**

3,910,701 A 10/1975 Henderson et al.  
3,954,560 A 5/1976 Delafosse et al.  
3,963,327 A 6/1976 Poirier  
(Continued)

(75) Inventor: **Zhi Xu**, St. Louis, MO (US)

**FOREIGN PATENT DOCUMENTS**

(73) Assignee: **St. Louis Medical Devices, Inc.**,  
Sunnyvale, CA (US)

CN 1192665 9/1998  
CN 2694097 Y 4/2005  
(Continued)

(\*) Notice: Subject to any disclaimer, the term of this  
patent is extended or adjusted under 35  
U.S.C. 154(b) by 520 days.

**OTHER PUBLICATIONS**

(21) Appl. No.: **13/610,140**

(22) Filed: **Sep. 11, 2012**

(65) **Prior Publication Data**

US 2013/0006070 A1 Jan. 3, 2013

**Related U.S. Application Data**

(62) Division of application No. 12/425,535, filed on Apr.  
17, 2009.

(60) Provisional application No. 61/089,152, filed on Aug.  
15, 2008, provisional application No. 61/055,303,  
filed on May 22, 2008.

Office Action for EP Application 08830786.3 dated Jan. 10, 2014.  
Office Action for CN Application 201210419849.3 dated Jan. 6,  
2014.  
Office Action for CN Application 201210420843.8 dated Feb. 17,  
2014.  
Office Action for CN Application 201210419740.X dated Feb. 28,  
2014.  
Office Action for EP Application 09751083.8 dated Mar. 28, 2014.  
Office Action for CN Application 201080022242.0 dated Mar. 12,  
2014.

(Continued)

(51) **Int. Cl.**  
**A61B 5/1455** (2006.01)  
**A61B 5/145** (2006.01)  
**A61B 5/00** (2006.01)

(52) **U.S. Cl.**  
CPC ..... **A61B 5/14532** (2013.01); **A61B 5/1455**  
(2013.01); **A61B 5/6826** (2013.01); **A61B**  
**5/6838** (2013.01)

(58) **Field of Classification Search**  
CPC ..... A61B 5/14528  
USPC ..... 600/309, 310, 316, 322, 323, 335, 336  
See application file for complete search history.

(56) **References Cited**

**U.S. PATENT DOCUMENTS**

2,441,343 A 5/1948 Becker  
3,621,268 A 11/1971 Friedrich et al.

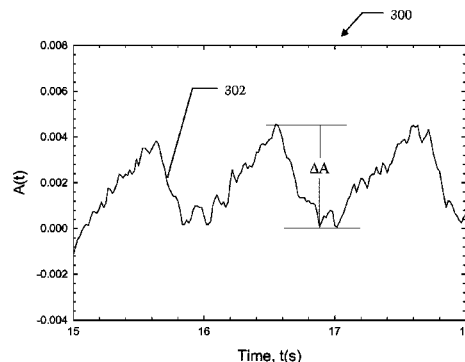
*Primary Examiner* — Michael C Stout

(74) *Attorney, Agent, or Firm* — Haverstock & Owens  
LLP

(57) **ABSTRACT**

Systems and methods are disclosed for non-invasively measuring blood glucose levels in a biological sample based on spectral data. This includes utilizing at least one light source configured to strike a target area of a sample, utilizing at least one light filter positioned to receive light transmitted through the target area from the at least one light source, utilizing at least one light detector positioned to receive light from the at least one light source and filtered by the at least one light filter, and to generate an output signal, having a time dependent current, which is indicative of the power of light detected, receiving the output signal from the at least one light detector with a processor and based on the received output signal, calculating the attenuation attributable to blood in a sample with a signal-to-noise ratio of at least 20-to-1; and determining a blood glucose level.

**6 Claims, 4 Drawing Sheets**



(56)	<b>References Cited</b>			7,254,432 B2	8/2007	Fine	
	U.S. PATENT DOCUMENTS			7,266,400 B2	9/2007	Fine et al.	
				7,409,239 B2	8/2008	Chung et al.	
				7,424,317 B2	9/2008	Parker et al.	
				7,436,511 B2 *	10/2008	Ruchti	G01J 3/28 356/319
4,014,321 A	3/1977	March		7,809,418 B2	10/2010	Xu	
4,632,559 A	12/1986	Brunsting		7,961,305 B2	6/2011	Xu et al.	
4,655,225 A	4/1987	Dahne et al.		8,272,771 B2	9/2012	Arai	
4,781,195 A	11/1988	Martin		8,340,738 B2	12/2012	Xu	
4,962,311 A	10/1990	Poysel et al.		2001/0030742 A1	10/2001	Kramer et al.	
4,997,769 A	3/1991	Lundsgaard		2001/0039376 A1	11/2001	Steuer et al.	
5,009,230 A	4/1991	Hutchinson		2001/0047137 A1 *	11/2001	Moreno	A61B 5/0075 600/475
5,028,787 A	7/1991	Rosenthal et al.		2002/0010563 A1	1/2002	Ratteree et al.	
5,077,476 A	12/1991	Rosenthal		2002/0016534 A1	2/2002	Trepagnier et al.	
5,086,229 A	2/1992	Rosenthal et al.		2002/0019055 A1	2/2002	Brown et al.	
5,112,124 A	5/1992	Harjunmaa et al.		2002/0038080 A1 *	3/2002	Makarewicz	A61B 5/14532 600/322
5,137,023 A	8/1992	Mendelson et al.		2002/0091324 A1	7/2002	Kollias et al.	
5,183,042 A	2/1993	Harjunmaa et al.		2002/0161289 A1	10/2002	Hopkins et al.	
5,222,496 A	6/1993	Clarke et al.		2002/0167704 A1	11/2002	Kleinhaus et al.	
5,255,171 A	10/1993	Clark		2003/0004423 A1	1/2003	Lavie et al.	
5,282,473 A	2/1994	Braig et al.		2003/0023152 A1	1/2003	Abbink et al.	
5,361,758 A	11/1994	Hall et al.		2003/0055325 A1 *	3/2003	Weber	A61B 5/024 600/323
5,423,983 A	6/1995	Chiang		2003/0078504 A1	4/2003	Rowe	
5,501,648 A *	3/1996	Grigoriev	B62M 1/12 482/57	2004/0015734 A1	1/2004	Rahman	
5,515,847 A	5/1996	Braig et al.		2004/0087844 A1	5/2004	Yen	
5,522,388 A	6/1996	Ishikawa et al.		2004/0106163 A1	6/2004	Workman et al.	
5,529,065 A	6/1996	Tsuchiya		2004/0127779 A1	7/2004	Steuer et al.	
5,535,743 A	7/1996	Backhaus et al.		2004/0181132 A1	9/2004	Rosenthal	
5,553,613 A	9/1996	Parker		2004/0204865 A1	10/2004	Lee et al.	
5,576,544 A	11/1996	Rosenthal		2004/0225205 A1	11/2004	Fine et al.	
5,615,672 A	4/1997	Braig et al.		2004/0225206 A1	11/2004	Kouchnir	
5,615,673 A	4/1997	Berger et al.		2005/0131286 A1	6/2005	Parker et al.	
5,666,956 A	9/1997	Buchert		2005/0197790 A1	9/2005	Sterling et al.	
5,671,301 A	9/1997	Kupersmidt		2005/0261560 A1	11/2005	Ridder et al.	
5,703,364 A	12/1997	Rosenthal		2005/0272987 A1	12/2005	Kiani-Azarbayjany et al.	
5,743,262 A	4/1998	Lepper, Jr. et al.		2005/0276072 A1	12/2005	Hayashi et al.	
5,910,109 A	6/1999	Peters et al.		2006/0002598 A1	1/2006	Rowe et al.	
6,025,597 A	2/2000	Sterling et al.		2006/0009685 A1	1/2006	Finarov et al.	
6,043,492 A	3/2000	Lee et al.		2006/0058622 A1	3/2006	Tearney et al.	
6,064,898 A	5/2000	Aldrich		2006/0063983 A1	3/2006	Yamakoshi	
6,067,463 A	5/2000	Jeng et al.		2006/0092643 A1	5/2006	Wong et al.	
6,097,975 A	8/2000	Petrovsky et al.		2006/0129040 A1	6/2006	Fine et al.	
6,134,458 A	10/2000	Rosenthal		2006/0152726 A1	7/2006	Larsen et al.	
6,151,517 A	11/2000	Honigs et al.		2006/0200014 A1	9/2006	Fine et al.	
6,167,290 A	12/2000	Yang et al.		2006/0224057 A1	10/2006	Burd et al.	
6,181,958 B1	1/2001	Steuer et al.		2006/0226992 A1	10/2006	Al-Ali et al.	
6,205,354 B1	3/2001	Gellermann et al.		2006/0234386 A1 *	10/2006	Burns	A61B 5/1455 436/94
6,208,788 B1	3/2001	Nosov		2006/0250676 A1	11/2006	Hagood	
6,304,767 B1	10/2001	Soller et al.		2006/0258918 A1	11/2006	Burd et al.	
6,312,393 B1	11/2001	Abreu		2006/0264719 A1	11/2006	Schurman et al.	
6,337,564 B2	1/2002	Manzini et al.		2007/0049811 A1	3/2007	Kobayashi et al.	
6,403,944 B1	6/2002	MacKenzie et al.		2007/0078312 A1	4/2007	Fine et al.	
6,421,548 B1	7/2002	Berman et al.		2007/0112258 A1 *	5/2007	Soyemi	A61B 5/0059 600/310
6,424,848 B1	7/2002	Berman et al.		2007/0149869 A1	6/2007	Yen	
6,424,849 B1	7/2002	Berman et al.		2008/0027297 A1	1/2008	Yamakoshi	
6,424,851 B1	7/2002	Berman et al.		2008/0144004 A1	6/2008	Rosenthal	
6,430,424 B1	8/2002	Berman et al.		2008/0194014 A1	8/2008	Young et al.	
6,445,938 B1	9/2002	Berman et al.		2008/0266900 A1	10/2008	Harbers et al.	
6,522,903 B1	2/2003	Berman et al.		2009/0059586 A1	3/2009	Livesay et al.	
6,574,490 B2	6/2003	Abbink et al.		2009/0079964 A1	3/2009	Xu	
6,671,528 B2	12/2003	Steuer et al.		2009/0105565 A1	4/2009	Xu	
6,684,099 B2	1/2004	Ridder et al.		2009/0116017 A1	5/2009	Xu et al.	
6,723,048 B2	4/2004	Fuller		2009/0196025 A1	8/2009	Joseph et al.	
6,731,963 B2	5/2004	Finarov et al.		2009/0247843 A1	10/2009	Xu	
6,775,564 B1	8/2004	Peters et al.		2009/0270700 A1	10/2009	Van Herpen et al.	
6,804,002 B2	10/2004	Fine et al.		2009/0292186 A1	11/2009	Xu	
6,833,540 B2	12/2004	MacKenzie et al.		2010/0026995 A1	2/2010	Merritt et al.	
6,865,408 B1	3/2005	Abbink et al.		2010/0252721 A1	10/2010	Xu	
6,873,865 B2	3/2005	Steuer et al.		2013/0006071 A1	1/2013	Xu	
6,958,039 B2	10/2005	Burd et al.		2013/0006072 A1	1/2013	Xu	
6,968,222 B2	11/2005	Burd et al.		2013/0006073 A1	1/2013	Xu	
6,990,365 B1	1/2006	Parker et al.					
6,993,372 B2	1/2006	Fine et al.					
7,039,446 B2 *	5/2006	Ruchti	A61B 5/0071 600/310				
7,039,447 B2	5/2006	Berman et al.					
7,043,289 B2	5/2006	Fine et al.					
7,107,087 B2	9/2006	Hwang et al.					
7,133,711 B2	11/2006	Chernoguz et al.					

(56)

**References Cited**

## U.S. PATENT DOCUMENTS

## FOREIGN PATENT DOCUMENTS

CN	1932840	A	3/2007
EP	0319159	A1	6/1989
EP	0781527	A1	7/1997
EP	01094745	A1	5/2001
EP	1281370	A2	2/2003
EP	1300712	A2	4/2003
EP	830582	B1	8/2005
GB	810256	A	3/1959
JP	56-156138	A	12/1981
JP	02-191434	A	7/1990
JP	07-088105	A	4/1995
JP	720551	U	4/1995
JP	9010238		1/1997
JP	H0956702	A	3/1997
JP	11037931	A	2/1999
JP	11-178813	A	7/1999
JP	2000083933	A	3/2000
JP	2003245265	A	9/2003
JP	2004267613	A	9/2004
JP	2004286475	A	10/2004
JP	2004290544	A	10/2004
JP	2004538054	A	12/2004
JP	2005-283563	A	10/2005
JP	2007185348	A	7/2007
JP	2009545344	A	12/2009
RU	2050545	C1	12/1995
RU	2188425	C2	8/2002
RU	2198402	C2	2/2003
SU	1193541	A1	11/1985
WO	90/13092	A1	11/1990
WO	9115991	A1	10/1991
WO	9115992	A1	10/1991
WO	92/00513	A1	1/1992
WO	9300856	A1	1/1993
WO	93/06774	A1	4/1993
WO	9316629	A1	9/1993
WO	9413199	A1	6/1994
WO	9416614	A1	8/1994
WO	95/05599	A1	2/1995
WO	9531930	A1	11/1995
WO	9604840	A1	2/1996
WO	9617546	A1	6/1996
WO	96/39926		12/1996
WO	96/41151	A1	12/1996
WO	9639927	A1	12/1996
WO	9803847	A2	1/1998
WO	9836681	A1	8/1998
WO	99/16136	A1	4/1999
WO	9939631	A1	8/1999
WO	0001294	A1	1/2000
WO	0016688	A1	3/2000
WO	01/16578	A1	3/2001
WO	0193755	A1	12/2001
WO	0196872	A2	12/2001
WO	02082990	A1	10/2002
WO	03010510	A2	2/2003
WO	03/077756	A1	9/2003
WO	03079900	A1	10/2003
WO	2005045377	A2	5/2005
WO	2006086566	A2	8/2006
WO	2006094109	A1	9/2006
WO	2007122557	A2	11/2007
WO	2008/039195	A1	4/2008
WO	2009/035669	A1	3/2009
WO	2009/045492	A1	4/2009
WO	2009/120600	A2	10/2009
WO	2009/142853	A1	11/2009
WO	2010017238	A1	2/2010
WO	2010114736		10/2010

## OTHER PUBLICATIONS

International Preliminary Report on Patentability (Chapter II) for PCT/US2008/011438 dated Jun. 18, 2010.

International Preliminary Report on Patentability (Chapter II) for PCT/US2009/037805 dated Dec. 14, 2010.

International Preliminary Report on Patentability (Chapter II) for PCT/US2009/040942 dated Dec. 13, 2010.

International Search Report and Written Opinion for PCT/US2010/028255 dated May 19, 2010.

International Search Report for PCT/US2008/010670 dated Nov. 21, 2008.

International Search Report for PCT/US2008/011438 dated Dec. 9, 2008.

Office Action for CN Application 200980126116.7 dated Jun. 4, 2012.

Office Action for RU Application 2010114587 dated Jun. 22, 2012.

Office Action for RU Application 2010117396 dated Jun. 18, 2012.

Office Action for U.S. Appl. No. 12/209,807 dated Sep. 17, 2010.

Office Action for U.S. Appl. No. 12/256,028 dated May 24, 2010.

Office Action for U.S. Appl. No. 12/256,028 dated Sep. 15, 2010.

Office Action for U.S. Appl. No. 12/407,999 dated Apr. 6, 2012.

Office Action for U.S. Appl. No. 12/729,886 dated Oct. 2, 2012.

Office Action for U.S. Appl. No. 12/209,807 dated May 17, 2010.

Wagner et al., "Invasiveness as a Barrier to Self-Monitoring of Blood Glucose in Diabetes", Diabetes Technology & Therapeutics, Aug. 1, 2005.

Web Page Document entitled [http://www.orsense.com/Diabetes\\_Monitoring](http://www.orsense.com/Diabetes_Monitoring) dated Aug. 9, 2007.

Office Action for CN Application 200880114960.3 dated Jan. 29, 2013.

Office Action for CN Application 200980126116.7 dated Feb. 16, 2013.

Office Action for JP Application 2010-524873 dated Dec. 25, 2012.

Office Action for JP Application 2010-527994 dated Dec. 25, 2012.

Office Action for U.S. Appl. No. 12/407,999 dated Nov. 21, 2012.

Office Action for U.S. Appl. No. 12/425,535 dated May 16, 2012.

Office Action for U.S. Appl. No. 12/729,886 dated Mar. 12, 2013.

Office Action for U.S. Appl. No. 12/425,535 dated Mar. 22, 2012.

Extended European Search Report for EP Application 08830786.3 dated Apr. 22, 2013.

Office Action for RU Application 2010152373 dated Mar. 26, 2013.

Office Action for CN Application 201080022242.0 dated Jul. 4, 2013.

Extended European Search Report for EP Application 09751083.8 dated Jul. 26, 2013.

Office Action for AU Application 2010232841 dated Aug. 13, 2013.

Office Action for JP Application 2011-501936 dated Jun. 25, 2013.

Examiner's Decision of Rejection for JP Application 2010-527994 dated Dec. 10, 2103.

Office Action for AU Application 2008299938 dated Sep. 13, 2013.

Office Action for CN Application 200980126116.7 dated Oct. 17, 2013.

Office Action for JP Application 2010-524873 dated Nov. 19, 2013.

Office Action for JP Application 2011-510533 dated Dec. 3, 2013.

Office Action for U.S. Appl. No. 12/407,999 dated Oct. 10, 2013.

Office Action for CA Application 2700996 dated Jul. 30, 2014.

Office Action for JP Application 2011-501936 dated Aug. 5, 2014.

Office Action for RU Application 2010114587 dated Mar. 25, 2014.

Office Action for U.S. Appl. No. 12/407,999 dated Nov. 20, 2014.

Office Action for EP Application 9751083.8 dated Jan. 26, 2016.

Office Action for CN Application 201310489245.0 dated Jan. 25, 2016.

Office Action for JP Application 2014-245584 dated Jan. 5, 2016, includes English translation.

Office Action for CA Application 2700996 dated Aug. 7, 2015.

Office Action for CA Application 2699626 dated Feb. 27, 2015.

Office Action for CN Application 200880114960.3 dated Feb. 17, 2015.

Office Action for CN Application 201210420830.0 dated Jan. 5, 2015.

Office Action for CN Application 201310489245.0 dated Feb. 26, 2015.

Office Action for EP Application 09751083.8 dated Feb. 24, 2015.

(56)

**References Cited**

OTHER PUBLICATIONS

Office Action for JP Application 2012-503498 dated Mar. 31, 2015.  
Office Action for RU Application 2011144084 dated Apr. 17, 2015.  
Office Action for U.S. Appl. No. 13/610,387 dated Jan. 22, 2015.  
Office Action for U.S. Appl. No. 13/610,423 dated Jan. 22, 2015.  
Office Action for U.S. Appl. No. 13/610,256 dated Jan. 21, 2015.  
Office Action for U.S. Appl. No. 12/407,999 dated Jan. 5, 2016.  
Extended European Search Report for EP Application 088360102 dated Mar. 8, 2016.  
Office Action for U.S. Appl. No. 13/610,423 dated Mar. 24, 2016.  
Office Action for CA Application 2789658 dated Mar. 30, 2016.  
Office Action for U.S. Appl. No. 13/610,342 dated Apr. 14, 2016.  
Office Action for U.S. Appl. No. 13/610,387 dated Apr. 14, 2016.  
Office Action for U.S. Appl. No. 13/610,256 dated May 20, 2016.

\* cited by examiner

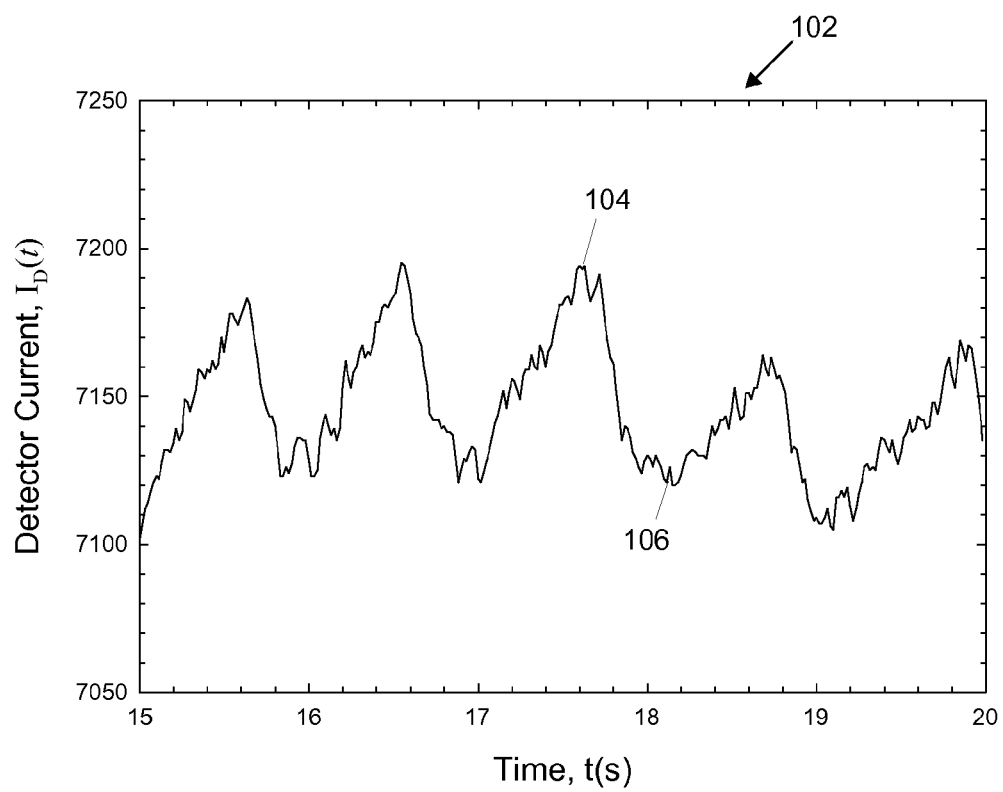
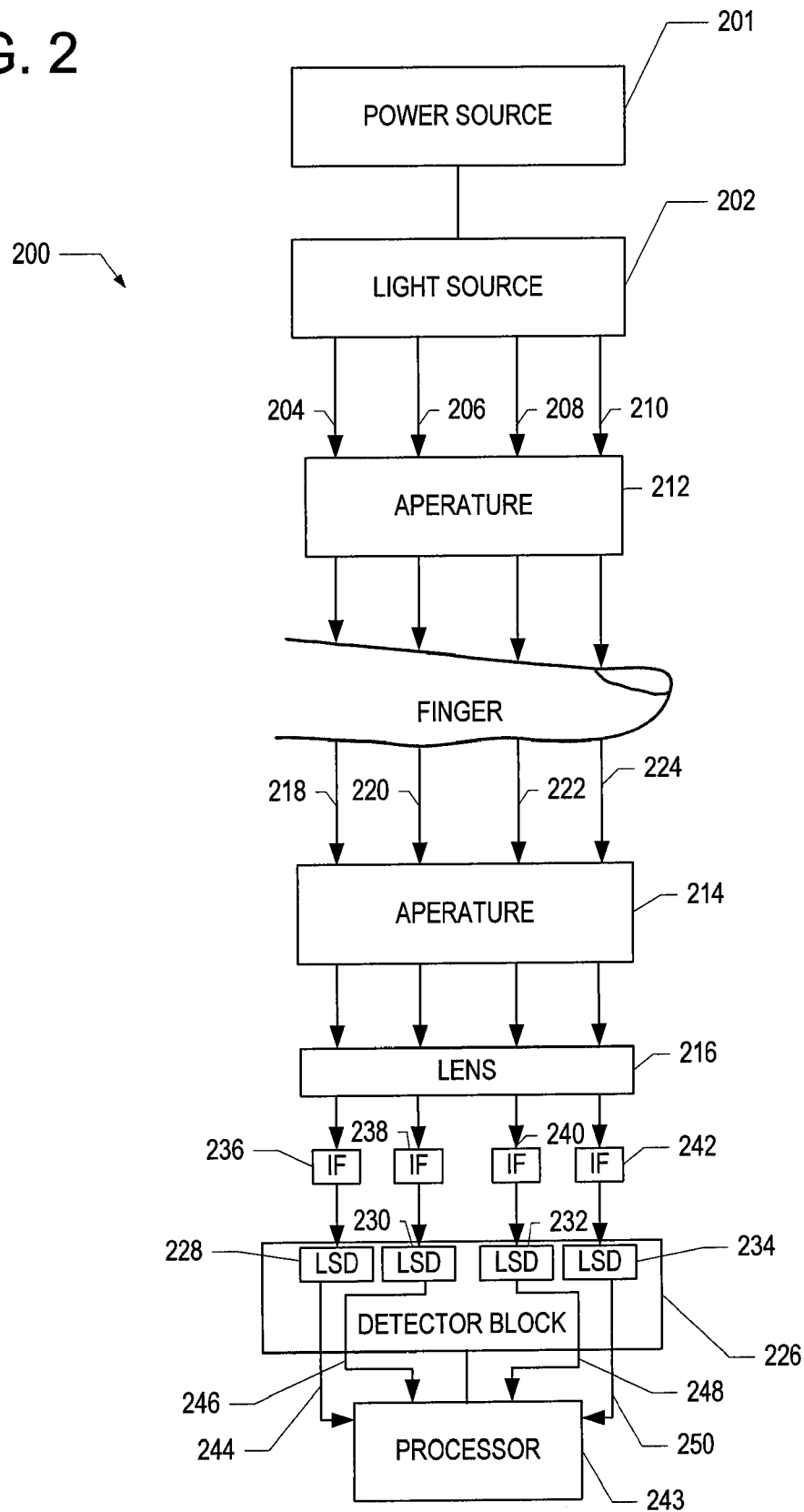


FIG. 1

FIG. 2



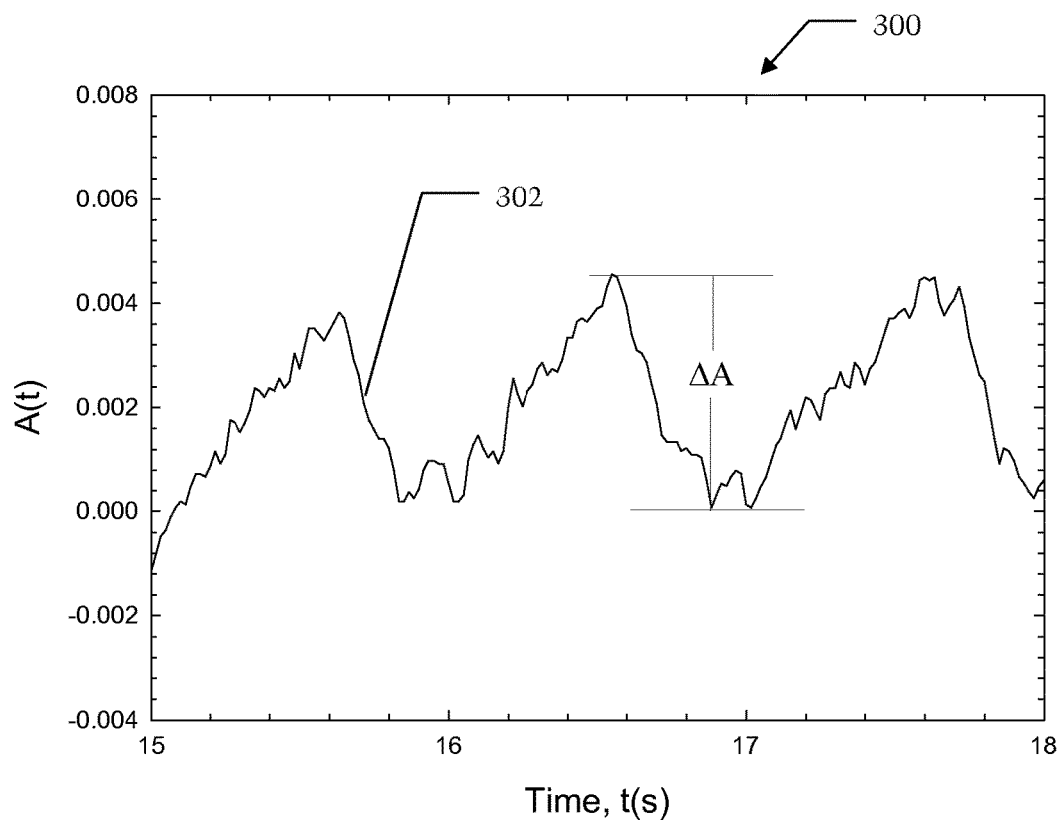


FIG. 3

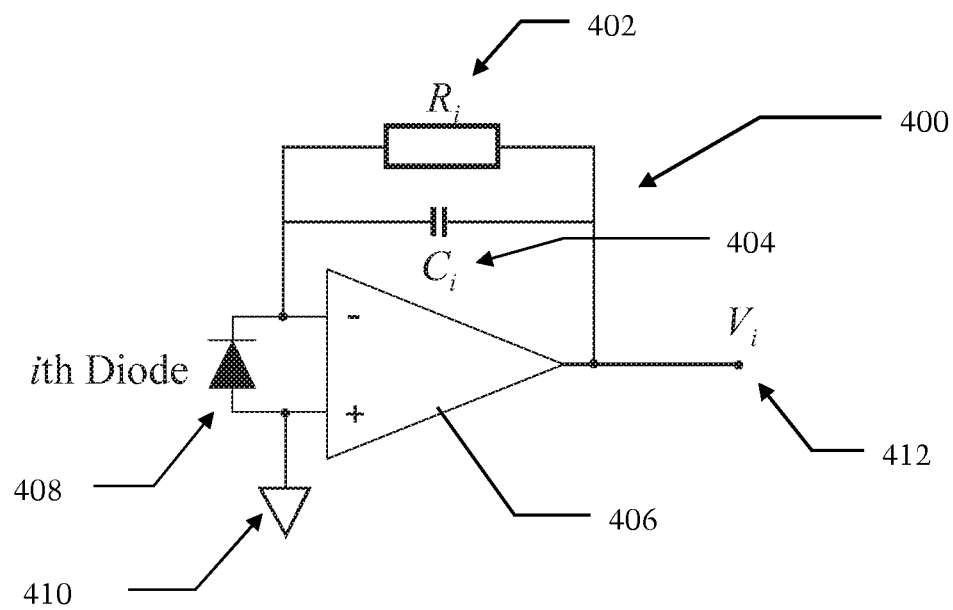


FIG. 4

1

# **METHOD AND SYSTEM FOR NON-INVASIVE OPTICAL BLOOD GLUCOSE DETECTION UTILIZING SPECTRAL DATA ANALYSIS**

## **CROSS-REFERENCE TO RELATED APPLICATION**

This application is a divisional of prior U.S. patent application Ser. No. 12/425,535, filed Apr. 17, 2009, which is hereby incorporated herein by reference in its entirety, and also claims priority to U.S. Provisional Patent Application Ser. No. 61/055,303, filed on May 22, 2008, the disclosure of which is incorporated herein by reference, and also claims priority to U.S. Provisional Patent Application Ser. No. 61/089,152, filed on Aug. 15, 2008, the disclosure of which is incorporated herein by reference.

## **BACKGROUND OF THE INVENTION**

Diabetes is a chronic disease that, when not controlled, over time leads to serious damage to many of the body's systems, including the nerves, blood vessels, eyes, kidneys and heart. The National Institute of Diabetes and Digestive and Kidney Diseases (NIDDK) estimates that 23.6 million people, or 7.8 percent of the population in the United States, had diabetes in 2007. Globally, the World Health Organization (WHO) estimates that more than 180 million people have diabetes, a number they expect to increase to 366 million by 2030, with 30.3 million in the United States. According to the WHO, an estimated 1.1 million people died from diabetes in 2005. They project that diabetes deaths will increase by more than 50% between 2006 and 2015 overall and by more than 80% in upper-middle income countries.

The economic burden from diabetes for individuals and society as a whole is substantial. According to the American Diabetes Association, the total annual economic cost of diabetes was estimated to be \$174 billion in the United States in 2007. This is an increase of \$42 billion since 2002. This 32% increase means the dollar amount has risen over \$8 billion more each year.

A vital element of diabetes management is the self-monitoring of blood glucose (SMBG) concentration by diabetics in the home environment. By testing blood glucose levels often, diabetics can better manage medication, diet, and exercise to maintain control and prevent the long-term negative health outcomes. In fact, the Diabetes Control and Complications Trial (DCCT), which followed 1,441 diabetics for several years, showed that those following an intensive-control program with multiple blood sugar tests each day, as compared with the standard-treatment group, had only one-fourth as many people develop diabetic eye disease, half as many develop kidney disease, one-third as many develop nerve disease, and far fewer people who already had early forms of these three complications got worse.

However, current monitoring techniques discourage regular use due to the inconvenient and painful nature of drawing blood through the skin prior to analysis, which causes many diabetics to not be as diligent as they should be for good blood glucose control. As a result, non-invasive measurement of glucose concentration is a desirable and beneficial development for the management of diabetes. A non-invasive monitor will make testing multiple times each day pain-free and more palatable for children with diabetes. According to a study published in 2005 (J. Wagner, C. Malchoff, and G. Abbott, *Diabetes Technology & Therapeu-*

2

tics, 7(4) 2005, 612-619), people with diabetes would perform SMBG more frequently and have improved quality of life with a non-invasive blood glucose monitoring device.

There exist a number of non-invasive approaches for blood glucose determination. One technique of non-invasive blood chemical detection involves collecting and analyzing light spectra data.

Extracting information about blood characteristics, such as glucose concentration from spectral or other data obtained from spectroscopy, is a complex problem due to the presence of components (e.g., skin, fat, muscle, bone, interstitial fluid) other than blood in the area that is being sensed. Such other components can influence these signals in such a way as to alter the reading. In particular, the resulting signal may be much larger in magnitude than the portion of the signal that corresponds to blood, and therefore limits the ability to accurately extract blood characteristics information.

The present invention is directed to overcoming one or more of the problems set forth above.

## **SUMMARY OF INVENTION**

In an aspect of the present invention, a system for detecting glucose in a biological sample is disclosed. This system includes at least one light source configured to strike a target area of a sample, at least one light filter positioned to receive light transmitted through the target area of the sample from the at least one light source, at least one light detector positioned to receive light from the at least one light source and filtered by the at least one light filter, and to generate an output signal, having a time dependent current, which is indicative of the power of light detected, and a processor configured to receive the output signal from the at least one light detector and based on the received output signal, calculate the attenuation attributable to blood in a sample present in the target area with a signal-to-noise ratio of at least 20-to-1, and based on the calculated attenuation, determine a blood glucose level associated with a sample present in the target area.

In yet another aspect of the present invention, a method for detecting glucose in a biological sample is disclosed. The method includes utilizing at least one light source configured to strike a target area of a sample, utilizing at least one light filter positioned to receive light transmitted through the target area of the sample from the at least one light source, utilizing at least one light detector positioned to receive light from the at least one light source and filtered by the at least one light filter, and to generate an output signal, having a time dependent current, which is indicative of the power of light detected, receiving the output signal from the at least one light detector with a processor and based on the received output signal, calculating the attenuation attributable to blood in a sample present in the target area with a signal-to-noise ratio of at least 20-to-1; and determining a blood glucose level associated with a sample present in the target area based on the calculated attenuation with the processor.

These are merely some of the innumerable aspects of the present invention and should not be deemed an all-inclusive listing of the innumerable aspects associated with the present invention.

## **BRIEF DESCRIPTION OF THE DRAWINGS**

For a better understanding of the present invention, reference may be made to accompanying drawings, in which:

3

FIG. 1 illustrates a plot of a pulse wave corresponding to light absorption of arterial blood, according to exemplary embodiments;

FIG. 2 illustrates an exemplary system for obtaining spectral data;

FIG. 3 illustrates a plot of  $A(t)$ , calculated according to Equation (9) using data in FIG. 1; and

FIG. 4 is a basic illustrative schematic of a preamplifier circuit that converts photocurrent into voltage prior to digitization.

#### DETAILED DESCRIPTION OF THE INVENTION

In the following detailed description, numerous exemplary specific details are set forth in order to provide a thorough understanding of the invention. However, it will be understood by those skilled in the art that the present invention may be practiced without these specific details, or with various modifications of the details. In other instances, well known methods, procedures, and components have not been described in detail so as not to obscure the present invention.

Optical spectroscopy can be used to determine the amount of light absorbed and scattered, i.e., attenuated, by a biological sample such as a human finger. By measuring the amount of light absorbed by the sample, it is possible to determine glucose, cholesterol, and hemoglobin levels of a subject non-invasively. Fingertip measurements are usually preferred because of the large concentration of capillaries in the fingertip and because of the conversion of arterial blood into venous blood that occurs in the fingertip. However, the techniques of the present invention are not limited to use with a fingertip. For example, the biological sample could be a human earlobe.

When light is transmitted through a biological sample, such as a human finger, the light is attenuated by various components of the finger including skin, muscle, bone, fat, interstitial fluid and blood. It has been observed, however, that light attenuation by a human finger exhibits a small cyclic pattern that corresponds to a heartbeat. It is believed that this cyclic pattern will be present in measurements of many other human body parts, the earlobe being one of many examples.

FIG. 1 depicts a plot 102 of a detector photocurrent,  $I_D(t)$ , that corresponds to the power of light received by a detector after the light has passed through a subject's finger. As can be seen, the detector photocurrent exhibits a cyclic pattern. This cyclic pattern is due to the subject's heartbeat, which cyclically increases and decreases the quantity of blood in the subject's capillaries (or other structures). Although the magnitude of the cyclic pattern is small in comparison to the total photocurrent generated by the detector, considerable information can be extracted from the cyclic pattern of the plot 102. For example, assuming that the person's heart rate is sixty beats per minute, the time between the start of any pulse beat and the end of that pulse beat is one second. During this one-second period, the photocurrent will have a maximum or peak reading 104 and minimum or valley reading 106. The peak reading 104 of the plot corresponds to when there is a minimum amount of blood in the capillaries, and the valley reading 106 corresponds to when there is a maximum amount of blood in the capillaries. By using information provided by the peak and valley of the cyclic plot, the optical absorption and scattering by major finger constituents that are not in the capillaries, such as skin, fat, bones, muscle and interstitial fluids, are excluded.

4

These major constituents that are not in the capillaries are excluded because they are not likely to change during the time interval of one heartbeat. In other words, the light that is absorbed and scattered, i.e., attenuated, by the blood can be detected based on the peaks and valleys of the plot 102.

Assuming that the peak of the cyclic photocurrent generated by the light-sensing device is  $I_P$ , the adjacent valley of the cyclic photocurrent is  $I_V$ , and the photocurrent generated by the light-sensing device without a human finger is  $I_0$ , the transmittances corresponding to the peak and valley photocurrents can be defined as:

$$T_V = \frac{I_V}{I_0} \quad (1)$$

and

$$T_P = \frac{I_P}{I_0} \quad (2)$$

The corresponding peak and valley absorbance are:

$$A_V = -\log(T_V) \quad (3)$$

and

$$A_P = -\log(T_P) \quad (4)$$

The difference between  $A_V$  and  $A_P$  represents the light absorption and scattering by the blood in the finger, excluding non-blood constituents:

$$\Delta A = A_V - A_P = \log\left(\frac{I_P}{I_V}\right) \quad (5)$$

As can be seen in the algorithm shown in Equation (5),  $\Delta A$  does not depend on  $I_0$ . Thus, calculating  $\Delta A$  does not require a determination of the current generated by the light-sensing device without a sample. Monitoring the photocurrent corresponding to light power transmitted through a sample is sufficient to calculate  $\Delta A$ .

FIG. 2 depicts a simplified block diagram of an exemplary apparatus for use in an exemplary embodiment. Optical measurement system, which is generally indicated by numeral 200, uses the "pulsatile" concept for determining an amount of light absorbed and scattered solely by the blood in a sample (a human finger in this exemplary embodiment). A power source 201, such as a battery, provides power to a light source 202 that generates a plurality of light beams 204, 206, 208, 210 that are directed toward the top of the finger of a subject. In an exemplary embodiment, each of the light beams 204, 206, 208, 210 have the same wavelength or a different wavelength range, typically within 800 nm to 1600 nm. Although the optical measurement system 200 is described herein as generating four (4) light beams, it is contemplated that the light source 202 can be altered to generate fewer light beams or additional light beams in other embodiments.

A first aperture 212 ensures that the light beams 204, 206, 208, 210 strike a target area of the finger. A second aperture 214 ensures that the portion of the light beams that are transmitted through the finger strike a lens 216. Light beams 204, 206, 208, 210 are attenuated by the finger and components of the optical measurement system 200, and, thus, attenuated light beams 218, 220, 222, 224 are emitted from the finger. The attenuated light beams 218, 220, 222, 224 strike the lens 216, and the lens 216 collects the attenuated

## 5

light beams **218, 220, 222, 224** so that they impinge more efficiently on a detector block **226**.

The detector block **226** is positioned directly under the lens **216** and comprises a plurality of light-sensing devices (LSD) **228, 230, 232, 234** such as an array of photodiodes. According to one aspect of the optical measurement system **200**, each of the light-sensing devices **228, 230, 232, 234** detects a specific wavelength of light as defined by corresponding interference filters (IF) **236, 238, 240, 242**, respectively. The interference filter transmits one or more spectral bands or lines of light, and blocks others.

Each of the light-sensing devices **228, 230, 232, 234** generates a corresponding photocurrent signal that is proportional to the power of the light received by the particular light sensing device. The photocurrent signal generated by the photodiode can be converted to another form of signal, such as an analog voltage signal or a digital signal. A processor **243** is coupled to the detector block **226** and is configured to calculate the change of photocurrent signals **244, 246, 248, 250**.

According to one aspect, the processor **243** executes an algorithm such as shown in the Equation (5) to calculate the change in the light absorption ( $\Delta A$ ) solely caused by the blood in the finger. Thereafter, this quantitative calculation of light absorption of the blood can be used to determine a characteristic of the blood. For example, by comparing the calculated light absorption value to predetermined values corresponding to different glucose levels stored in a memory (not shown), a blood-glucose level of the subject can be determined.

A difficulty associated with the finger based pulsatile detection methodology is low signal-to-noise (S/N) ratio, because the amplitude of cyclic pattern (i.e., the difference between peak and valley) is typically 1%-2% of the total photocurrent generated by the light power transmitted through the finger. To obtain a S/N ratio of 100:1 in the determination of  $\Delta A$ , the baseline noise of the device being used to measure the light absorption by the finger should not be larger than  $3.0 \times 10^{-5}$  in absorbance (peak to peak), within a 10 Hz bandwidth.

However, a  $3.0 \times 10^{-5}$  absorbance (peak to peak) baseline noise level within a 10 Hz bandwidth is difficult to obtain with the low light power levels that are used by some battery-powered hand held non-invasive blood chemical measurement devices. One solution involves data averaging. To increase the S/N ratio, the averaged value of  $\Delta A$ , as defined by the Equation below, is used in further calculation to extract blood glucose concentration:

$$\bar{\Delta A} = \sum_{j=1}^M \Delta A_j \quad (6)$$

In Equation (6), M is the number of heartbeats during the time interval of the pulsatile measurement. However, this approach requires long data acquisition time, due to the fact that the rate of heartbeat is in the order of one per second. For example, 25 seconds would be needed for increasing the S/N ratio by a factor of five, and 100 seconds would be needed for increasing the S/N ratio by a factor of ten. In comparison, current commercial blood drawing glucose meters can determine blood glucose level within 5 seconds. Furthermore, long detection time will significantly increase measurement errors due to finger movement, light power

## 6

drift, device temperature change, etc. Thus, there is a need for new techniques to measure blood glucose levels quickly and accurately.

Improving S/N Ratio by Standard Deviation

The time dependent detector photocurrent output,  $I_D(t)$ , shown in FIG. 1 can be expressed as the sum of a small time dependent cyclic photocurrent  $\Delta I(t)$ , corresponding to the heartbeat, a noise current  $n(t)$ , and a constant baseline photocurrent  $I_B$ :

$$I_D(t) = I_B + \Delta I(t) + n(t) \quad (7)$$

The above Equation can be re-arranged as:

$$\frac{I_D(t)}{I_B} = 1 + \frac{\Delta I(t) + n(t)}{I_B} \quad (8)$$

Applying common logarithm to both side of the Equation (8), one obtains:

$$A(t) = \log\left[\frac{I_D(t)}{I_B}\right] = \log\left(1 + \frac{\Delta I(t) + n(t)}{I_B}\right) \quad (9)$$

FIG. 3, which is generally indicated by numeral **300**, shows a typical A(t) plot **302**, calculated according to Equation (9) using data in FIG. 1. For a pulse function A(t) shown in FIG. 3, the following key relationship exists during the time interval of one heartbeat:

$$\sigma[A(t)] = k \Delta A \quad (10)$$

in which  $\sigma[A(t)]$  is the Standard Deviation of A(t), and k is a proportional constant.

Considering the fact that  $I_B$  is a constant and  $\sigma^2(\log I_B) = 0$ , one obtains:

$$\sigma[A(t)] = \sigma[\log I_D(t)] \quad (12)$$

Therefore, the peak-to-valley height of the A(t) plot during the time interval of one heartbeat can be obtained directly from the standard deviation of the logarithm of  $I_D(t)$ :

$$\Delta A = \frac{\sigma[A(t)]}{k} = \frac{\sigma[\log I_D(t)]}{k} \quad (13)$$

A major advantage of Equation (13) is that high S/N ratio can be achieved within short data acquisition time (approximately one second), as explained below.

In a finger based pulsatile measurement depicted by FIG. 2, the value of the sum,  $\Delta I(t) + n(t)$  is typically less than 2% of the large constant baseline photocurrent  $I_B$ . Therefore, Equation (9) can be approximated as:

$$A(t) = \log\left[\frac{I_D(t)}{I_B}\right] \approx \frac{1}{\ln 10} \frac{\Delta I(t) + n(t)}{I_B} \quad (14)$$

Similarly, the standard deviation of A(t) can be approximated as:

$$\sigma[A(t)] \approx \frac{1}{\ln 10} \frac{\sqrt{\sigma^2[\Delta I(t)] + \sigma^2[n(t)]}}{I_B} \quad (15)$$

Equation (15) demonstrates great noise reduction power of Equation (13). For example, for a relatively high baseline noise with the ratio

$$\rho = \frac{\sigma[n(t)]}{\sigma[\log \Delta I(t)]} = 0.1 \text{ (or 10\%)},$$

the contribution to  $\sigma[A(t)]$  from the baseline noise  $n(t)$  is estimated to be less than 0.005 (or 0.5%), corresponding to an increase in S/N ratio by a factor of 20 without increasing detection time. As such, dramatic noise reduction can be obtained without increasing the data acquisition time, and a finger based pulsatile measurement can be completed within the time interval of one heartbeat (which is approximately one second), and the requirement for the S/N ratio of 100 to 1 in determination of  $\Delta A$  can be satisfied using an optical system with a baseline noise of about  $6.0 \times 10^{-4}$  absorbance (peak to peak) within a 10 Hz bandwidth. It should be pointed out that when the baseline noise of an optical system is dominated by shot noise due to low light illumination power, a noise reduction by a factor of 20 equals an increasing in light illumination power by a factor of  $20^2=400$ .

This ability of obtaining higher S/N ratio within the very short data acquisition time, e.g., less than one second, will significantly reduce detection error caused by factors such as finger movement, temperature change, and light power drift during the measurement, and therefore dramatically improve the accuracy and reproducibility of the pulsatile detection methodology.

Furthermore, the value of  $k$  does not change with wavelength, because transmitted lights at all wavelengths have identical pulse shape due to the heartbeat. As a result, the constant  $k$  will be cancelled in data normalization discussed in next section, and  $\sigma[\log I_D(t)]$  will be used in further regression analysis to establish correlation between the optical measurement and blood glucose level. This will greatly simplify the data analysis process since  $\sigma[\log I_D(t)]$  involves only two standard math functions available in most popular spreadsheet programs such as Microsoft EXCEL®. EXCEL® is a federally registered trademark of Microsoft Corporation, having a place of business at One Microsoft Way, Redmond, Wash. 98052-6399.

#### Normalization

At each wavelength  $\lambda_i$ , the absorption  $\Delta A(\lambda_i)$  is linked to the increase of amount of blood ( $\Delta B$ ) in the optical sensing area of the fingertip due to the heartbeat by the following Equation:

$$\Delta A(\lambda_i) = \epsilon(C, \lambda_i, T) \Delta B \quad (16)$$

in which  $\epsilon(C, \lambda_i, T)$  is the absorption/scattering coefficient of blood at wavelength  $\lambda_i$ , finger temperature  $T$ , and blood glucose concentration  $C$ . It is well understood that the variable  $\Delta B$  differs from person to person, and may even change from day to day for the same person.

The uncertainty from the variable  $\Delta B$  can be cancelled by introducing the normalization factor  $Q_i(C, T)$  at each wavelength  $\lambda_i$ , as defined by the Equation below:

$$Q_i(C, T) = \frac{\Delta A(\lambda_i)}{\sum_{i=1}^N \Delta A(\lambda_i)} = \frac{\epsilon(C, \lambda_i, T)}{\sum_{i=1}^N \epsilon(C, \lambda_i, T)}, \quad (17)$$

in which  $N$  is total number of wavelength employed. Preferably,  $N$  typically ranges from twenty to thirty.

Based on Equations (13) and (17),  $Q_i(C, T)$  is linked to the detector photocurrent at each wavelength  $\lambda_i$ ,  $I_D(\lambda_i, t)$ , by the following Equation:

$$\begin{aligned} Q_i(C, T) &= \frac{\Delta A(\lambda_i)}{\sum_{i=1}^N \Delta A(\lambda_i)} \\ &= \frac{\sigma[\log I_D(\lambda_i, t)]/k}{\sum_{i=1}^N \sigma[\log I_D(\lambda_i, t)]/k} \\ &= \frac{\sigma[\log I_D(\lambda_i, t)]}{\sum_{i=1}^N \sigma[\log I_D(\lambda_i, t)]}, \end{aligned} \quad (18)$$

As shown by Equation (18), the constant  $k$  is cancelled and  $\sigma[\log I_D(t)]$  will be used in further regression analysis to establish correlation between the optical measurement and blood glucose level. This is possible because data are taken simultaneously from all detection channels.

A correlation between optical measurement and blood glucose concentration can be established according to the following Equation:

$$C_{optical} = \sum_{i=1}^N a_i(T) Q_i(C, T) \quad (19)$$

in which  $C_{optical}$  is the blood glucose concentration predicted by the optical measurement,  $Q_i(C, T)$  is defined by Equations (17) and (18), and  $a_i(T)$  is the temperature dependent regression coefficient corresponding to wavelength  $\lambda_i$ . The values of  $a_i(T)$  can be extracted using proper statistics methods such as Partial Least Squares (PLS) regression.

Equation (19) represents ideal cases when large number of calibrations can be made at different finger temperatures. In reality, frequently only a limited number of calibrations can be made (e.g., 15 to 20), and each may be taken at a different finger temperature. Under this condition, the finger temperature can be treated as an independent variable, and the above Equation can be approximated as:

$$C_{optical} = \sum_{i=1}^N b_i Q_i(C, T) + \eta T \quad (20)$$

in which  $b_i$  is the temperature independent regression coefficient corresponding to wavelength  $\lambda_i$ , and  $\eta$  is the regression coefficient for the finger temperature. The values of  $b_i$  and that of  $\eta$  can be extracted using proper statistics methods such as Partial Least Squares (PLS) regression.

#### Ratio Methodology

Alternatively, the uncertainty from the variable  $\Delta B$  can be cancelled by introducing a ratio factor  $Y_{ij}$  at wavelength  $\lambda_i$ :

$$\begin{aligned} Y_{ij}(C, T) &= \frac{\Delta A(\lambda_i)}{\Delta A(\lambda_j)} \\ &= \frac{\epsilon(C, \lambda_i, T)}{\epsilon(C, \lambda_j, T)} \end{aligned} \quad (21)$$

-continued

$$= \frac{\sigma[\log I_D(\lambda_i, t)]}{\sigma[\log I_D(\lambda_j, T)]},$$

in which j can be any number from 1 to N, assuming that the device collects signal at all N wavelengths.

Similar to the normalization algorithm discussed before, a correlation between optical measurement and blood glucose level can be established according to the following Equation:

$$C_{optical} = \sum_{i \neq j}^N f_i(T) Y_{ij}(C, T) \quad (22)$$

in which  $C_{optical}$  is the blood glucose concentration predicted by the optical measurement,  $Y_{ij}(C, T)$  is defined by Equation (21), and  $f_i(T)$  is the temperature dependent regression coefficient corresponding to wavelength  $\lambda_i$ . The value of  $f_i(T)$  can be obtained using statistics methods such as Partial Least Squares (PLS) regression.

Equation (22) represents ideal cases when large number of calibration can be made at different finger temperatures. In reality, frequently only limited number of calibration can be made (e.g., 15 to 20), and each may be taken at a different finger temperature. Under this condition, the finger temperature can be treated as an independent variable, and the above Equation can be approximated as:

$$C_{optical} = \sum_{i \neq j}^N h_i Y_{ij}(C, T) + \beta T \quad (23)$$

in which  $h_i$  is the temperature independent regression coefficient corresponding to wavelength  $\lambda_i$ , and  $\beta$  is the regression coefficient for the finger temperature. The values of  $h_i$  and that of  $\beta$  can be extracted using proper statistics methods such as Partial Least Squares (PLS) regression. Elimination of the Effect of Temperature Dependent Device Response

It is well understood that the detector sensitivity of a silicon photodiode detector is a function of wavelength and temperature. For the device configuration shown in FIG. 2, which is generally indicated by numeral 200, the light power received by ith silicon diode detector, corresponding to wavelength  $\lambda_i$ , is converted into a photocurrent according to the following Equation:

$$I_D(\lambda_i, t) = P(\lambda_i, t) S_0(\lambda_i) [1 + \gamma(\lambda_i)(T_{Di}(t) - 25^\circ \text{ C.})] \quad (24)$$

In the above Equation (24),  $P(\lambda_i, t)$  is the light power received by the detector,  $S_0(\lambda_i)$  is the photosensitivity of the detector at wavelength  $\lambda_i$  and  $25^\circ \text{ C.}$ ,  $\gamma(\lambda_i)$  is the temperature coefficient of the photosensitivity at wavelength  $\lambda_i$ , and  $T_{Di}(t)$  is the temperature of ith photodiode detector. The temperature coefficient  $\gamma(\lambda_i)$  varies with the wavelength. For example, for Hamamatsu S1337 series photodiode detectors,  $\gamma(\lambda_i)$  ranges from near zero at 900 nm to over 1.0%/° C. at 1100 nm. This imposes a potential problem for the device configuration show in FIG. 2, because it is very difficult to keep temperature of each individual diode detector constant in a handheld device used by a person with diabetes under a normal household/office environment.

This uncertainty due to the detector temperature  $T_{Di}(t)$  can be eliminated using the algorithm shown by Equations

(12) and (13). Applying common logarithm on both sides of the Equation (24), one obtains:

$$\log I_D(\lambda_i, t) = \log P(\lambda_i, t) + \log S_0(\lambda_i) + \log [1 + \gamma(\lambda_i)(T_{Di}(t) - 25^\circ \text{ C.})] \quad (25)$$

Considering the fact that  $S_0(\lambda_i)$  is a constant and that detector temperature  $T_{Di}(t)$  remains almost constant during the very short data acquisition time interval of approximately one second, one obtains:

$$\sigma[\log I_D(\lambda_i, t)] = \sigma[\log P(\lambda_i, t)] \quad (26)$$

As such, the uncertainty caused by detector temperature  $T_{Di}(t)$  is eliminated by the use of this standard deviation methodology.

Voltage Detection Mode

In the device configuration shown in FIG. 2, the photocurrent of ith photodiode detector  $I_D(\lambda_i, t)$  is typically converted into a voltage using a preamplifier before digitization. FIG. 4 shows the schematic circuit diagram of a typical preamplifier, which is generally indicated by numeral 400.

The output voltage 412 of ith preamplifier 400, in coupling with ith photodiode detector 408, can be expressed as:

$$V_i(t) = R_{Di} I_D(\lambda_i, t) = R_{Di} [1 + \chi_d(T_{Ri}(t) - 25^\circ \text{ C.})] I_D(\lambda_i, t) \quad (27)$$

In the above Equation (27),  $R_{Di}$  is the resistance value of feedback resistor 402 for ith preamplifier at  $25^\circ \text{ C.}$ ,  $\chi_d$  is the temperature coefficient of the resistor, and  $T_{Ri}(t)$  is the temperature of the resistor. Applying common logarithm to both side of the Equation (27), one obtains:

$$\log V_i(t) = \log R_{Di} + \log [1 + \chi_d(T_{Ri}(t) - 25^\circ \text{ C.})] + \log I_D(\lambda_i, t) \quad (28)$$

Considering the fact that  $R_{Di}$  is a constant and that the resistor temperature  $T_{Ri}(t)$  does not change during the very short data acquisition time interval of approximately one second, one obtains:

$$\sigma[\log V_i(t)] = \sigma[\log I_D(\lambda_i, t)] \quad (29)$$

Substituting Equation (26) into Equation (29), one obtains:

$$\sigma[\log V_i(t)] = \sigma[\log P(\lambda_i, t)] \quad (30)$$

As such, the uncertainty caused by resistor temperature  $T_{Ri}(t)$  is eliminated.

Under the voltage detection mode, the normalization factor in Equation (18) can be expressed as:

$$Q_i(C, T) = \frac{\sigma[\log V_i(t)]}{\sum_{i=1}^N \sigma[\log V_i(t)]} \quad (31)$$

The mathematic correlation between optical measurement and blood glucose concentration can then be established according to Equation (19) or Equation (20), under corresponding calibration conditions.

Similarly, the ratio factor defined by Equation (21) can be expressed as:

$$Y_{ij}(C, T) = \frac{\sigma[\log V_i(t)]}{\sigma[\log V_j(t)]} \quad (32)$$

The mathematic correlation between optical measurement and blood glucose concentration can then be established according to Equation (22) or Equation (23), under corre-

11

sponding calibration conditions. The schematic circuit diagram of a typical preamplifier **400** also includes a feedback capacitor **404**, an operational amplifier **406**, and a ground connection **410**.

Digitization

The voltage output **412** from the preamplifier **400** is usually digitized using an analog-to-digital convertor (ADC). The digitized signal is then sent to a computer for data analysis. The output of *i*th ADC, in communication with *i*th preamplifier that is in coupling with *i*th photodiode **408** collecting light power at wavelength  $\lambda_i$ , can be expressed by the following Equation:

$$(ADC)_i = (ADC)_{0i} + G_i \{ I_D(\lambda_i, t) + I_{Dark,i} R_i + A_{0i} \} \quad (33)$$

In the above Equation (33),  $(ADC)_{0i}$  is the offset of *i*th ADC,  $G_i$  is the nominal ADC Gain used during the detection,  $I_D(\lambda_i, t)$  is the photocurrent of *i*th photodiode detector,  $I_{Dark,i}$  is the dark current of *i*th photodiode detector,  $R_i = R_{0i} [1 + \gamma_i (T_{Ri}(t) - 25^\circ \text{C})]$  is the resistance of feedback resistor of *i*th preamplifier, and  $A_{0i}$  is the offset of *i*th preamplifier.

The contribution of the three factors,  $(ADC)_{0i}$ ,  $I_{Dark,i}$ , and  $A_{0i}$  can be removed by carrying out a dark measurement with the light source turned off right before or after the corresponding finger measurement. When the light source is turned off, the above Equation (33) becomes:

$$(ADC)_{Dark,i} = (ADC)_{0i} + G_i (I_{Dark,i} R_i + A_{0i}) \quad (34)$$

The difference between the two above Equations (33) and (34) reflects ADC output corresponding to the photocurrent:

$$\Delta(ADC)_i = (ADC)_i - (ADC)_{Dark,i} = G_i I_D(\lambda_i, t) R_i \quad (35)$$

Applying common logarithm to both side of the Equation (35), one obtains:

$$\log \Delta(ADC)_i = \log G_i + \log I_D(\lambda_i, t) + \log R_i \quad (36)$$

$G_i$  and  $R_i$  can be considered as constants as long as the time interval between the finger measurement and the dark measurement is short. As such, one obtains:

$$\sigma[\log \Delta(ADC)_i] = \sigma[\log I_D(\lambda_i, t)] \quad (37)$$

Substituting Equation (26) into Equation (37), one further obtains:

$$\sigma[\log \Delta(ADC)_i] = \sigma[\log P(\lambda_i, t)] \quad (38)$$

Based on Equation (37), the normalization factor defined by Equation (18) can be expressed as:

$$Q_i(C, T) = \frac{\sigma[\log \Delta(ADC)_i]}{\sum_{i=1}^N \sigma[\log \Delta(ADC)_i]} \quad (39)$$

The mathematic correlation between optical measurement and blood glucose concentration can then be established according to Equation (19) or (20), under corresponding calibration conditions.

Similar to normalization, the ratio factor defined by Equation (21) can be expressed as:

$$Y_{ij}(C, T) = \frac{\sigma[\log \Delta(ADC)_i]}{\sigma[\log \Delta(ADC)_j]} \quad (40)$$

12

The correlation between optical measurement and blood glucose concentration can then be established according to Equations (22) or (23), under corresponding calibration conditions.

Thus, there has been shown and described several embodiments of a novel invention. As is evident from the foregoing description, certain aspects of the present invention are not limited by the particular details of the examples illustrated herein, and it is therefore contemplated that other modifications and applications, or equivalents thereof, will occur to those skilled in the art. The terms "have," "having," "includes," "including," and similar terms as used in the foregoing specification are used in the sense of "optional" or "may include" and not as "required." Many changes, modifications, variations and other uses and applications of the present construction will, however, become apparent to those skilled in the art after considering the specification and the accompanying drawings. All such changes, modifications, variations and other uses and applications, which do not depart from the spirit and scope of the invention, are deemed to be covered by the invention, which is limited only by the claims that follow. It should be understood that the embodiments disclosed herein include any and all combinations of features described in any of the dependent claims.

The invention claimed is:

1. A system for detecting glucose in a biological sample, comprising:

at least one light source configured to generate one or more light beams having a wavelength in a range between 800 nm and 1600 nm to strike a target area of a biological sample;

at least one photocurrent signal generating light detector positioned to receive light from the at least one light source and to generate an output photocurrent signal, having a time dependent current, which is indicative of the power of light detected;

wherein the time dependent output current is a function of a baseline current, a noise current and a time dependent cyclic current corresponding to a heartbeat; and

a light absorbance change determining algorithm implemented processor programmed to calculate a change in a light absorption caused by blood in the biological sample and configured to receive the output photocurrent signal from the at least one photocurrent signal generating light detector and based on the received output photocurrent signal, calculate the attenuance attributable to blood in the biological sample present in the target area with a signal-to-noise ratio of at least 20-to-1, and based on the calculated attenuance, determine a blood glucose level associated with the biological sample present in the target area; and

wherein the calculated attenuance is based on an equation for improving a signal-to-noise ratio having at least in part on a standard deviation of a logarithm of the time dependent output current generated by the power of light from the same target area of the biological sample, wherein the equation comprises

$$\Delta A = \frac{\sigma[A(t)]}{k} = \frac{\sigma[\log I_D(t)]}{k},$$

where  $A(t)$  is the change in light absorption due to blood in the biological sample as a function of time,  $\Delta A$  is the peak-to-valley height of  $A(t)$  plot during the time interval of one heartbeat,  $I_D(t)$  is the time dependent detector current,

13

$\log I_D(t)$  is the logarithm of the time dependent detector current,  $k$  is the proportionality constant,  $\sigma[A(t)]$  is the standard deviation of  $A(t)$ , and  $\sigma[\log I_D(t)]$  represents the standard deviation of  $\log I_D(t)$ .

2. A system for detecting glucose in a biological sample, comprising

at least one light source configured to generate one or more light beams having a wavelength in a range between 800 nm and 1600 nm to strike a target area of a biological sample;

at least one photocurrent signal generating light detector positioned to receive light from the at least one light source and to generate an output photocurrent signal, having a time dependent current, which is indicative of the power of light detected;

wherein the time dependent output current is a function of a baseline current, a noise current and a time dependent cyclic current corresponding to a heartbeat; and

a light absorbance change determining algorithm implemented processor programmed to calculate a change in a light absorption caused by blood in the biological sample and configured to receive the output photocurrent signal from the at least one photocurrent signal generating light detector and based on the received photocurrent output signal, calculate attenuance attributable to blood in the biological sample present in the target area with a signal-to-noise ratio of at least 20-to-1, and based on the calculated attenuance, determine a blood glucose level associated with the biological sample present in the target area; and wherein the calculated attenuance is based on an equation for noise reduction having at least in part on an approximation of a standard deviation of a logarithm of the time dependent output current generated by the power of light from the same target area of the biological sample, wherein the equation comprises

$$\Delta A = \frac{\sigma[A(t)]}{k} = \frac{\sigma[\log I_D(t)]}{k},$$

where  $A(t)$  is the change in light absorption due to blood in the biological sample as a function of time,  $\Delta A$  is the peak-to-valley height of  $A(t)$  plot during the time interval of one heartbeat,  $I_D(t)$  is the time dependent detector current,  $\log I_D(t)$  is the logarithm of the time dependent detector current,  $k$  is the proportionality constant,  $\sigma[A(t)]$  is the standard deviation of  $A(t)$ , and  $\sigma[\log I_D(t)]$  represents the standard deviation of  $\log I_D(t)$ .

3. A system for detecting glucose in a biological sample, comprising:

at least one light source configured to generate one or more light beams having a wavelength in a range between 800 nm and 1600 nm to strike a target area of a biological sample;

at least one photocurrent signal generating light detector positioned to receive light from the at least one light source and to generate an output photocurrent signal, having a time dependent current, which is indicative of the power of light detected;

wherein the time dependent output current is a function of a baseline current, a noise current and a time dependent cyclic current corresponding to a heartbeat; and

a light absorbance change determining algorithm implemented processor programmed to calculate a change in a light absorption caused by blood in the biological

14

sample and configured to receive the output photocurrent signal from the at least one photocurrent signal generating light detector and based on the received output photocurrent signal, calculate attenuance attributable to blood in the biological sample present in the target area with a signal-to-noise ratio of at least 20-to-1, and based on the calculated attenuance, determine a blood glucose level associated with the biological sample present in the target area; and

wherein the light absorbance change determining algorithm implemented processor is configured to calculate a peak-to-valley height of the change in light absorption due to blood in the biological sample in relationship to time, which is a function of a standard deviation of a logarithm of the time dependent output current divided by a proportionality constant:

$$\Delta A = \frac{\sigma[A(t)]}{k} = \frac{\sigma[\log I_D(t)]}{k},$$

where  $A(t)$  is the change in light absorption due to blood in the biological sample as a function of time,  $\Delta A$  is the peak-to-valley height of  $A(t)$  plot during the time interval of one heartbeat,  $I_D(t)$  is the time dependent detector current,  $\log I_D(t)$  is the logarithm of the time dependent detector current,  $k$  is the proportionality constant,  $\sigma[A(t)]$  is the standard deviation of  $A(t)$ , and  $\sigma[\log I_D(t)]$  represents the standard deviation of  $\log I_D(t)$ .

4. A method for detecting glucose in a biological sample, comprising:

utilizing at least one light source configured to generate one or more light beams having a wavelength in a range between 800 nm and 1600 nm to strike a target area of a biological sample;

utilizing at least one photocurrent signal generating detector positioned to receive light from the at least one light source and to generate an output photocurrent signal, having a time dependent current, which is indicative of the power of light detected;

receiving the output photocurrent signal from the at least one photocurrent signal generating light detector with a light absorbance change determining algorithm implemented processor programmed to calculate a change in a light absorption caused by blood in the biological sample based on the received output photocurrent signal;

calculating the attenuance attributable to blood in the biological sample present in the target area with a signal-to-noise ratio of at least 20-to-1;

determining a blood glucose level associated with the biological sample present in the target area based on the calculated attenuance with the light absorbance change determining algorithm implemented processor; and

wherein the step of calculating the attenuance attributable to blood in the biological sample present in the target area is based on an equation for improving a signal-to-noise ratio having at least in part on a standard deviation of a logarithm of the time dependent output current generated by the light power from the same target area of the biological sample, wherein the equation comprises

$$\Delta A = \frac{\sigma[A(t)]}{k} = \frac{\sigma[\log I_D(t)]}{k},$$

15

where  $A(t)$  is the change in light absorption due to blood in the biological sample as a function of time,  $\Delta A$  is the peak-to-valley height of  $A(t)$  plot during the time interval of one heartbeat,  $I_D(t)$  is the time dependent detector current,  $\log I_D(t)$  is the logarithm of the time dependent detector current,  $k$  is the proportionality constant,  $\sigma[A(t)]$  is the standard deviation of  $A(t)$ , and  $\sigma[\log I_D(t)]$  represents the standard deviation of  $\log I_D(t)$ .

5. A method for detecting glucose in a biological sample, comprising:

utilizing at least one light source configured to generate one or more light beams having a wavelength in a range between 800 nm and 1600 nm to strike a target area of a biological sample;

utilizing at least one photocurrent signal generating light detector positioned to receive light from the at least one light source and to generate an output photocurrent signal, having a time dependent current, which is indicative of the power of light detected;

receiving the output photocurrent signal from the at least one photocurrent signal generating light detector with a light absorbance change determining algorithm implemented processor programmed to calculate a change in a light absorption caused by blood in the biological sample based on the received output photocurrent signal;

calculating the attenuance attributable to blood in a biological sample present in the target area with a signal-to-noise ratio of at least 20-to-1 utilizing a peak-to-valley height of the change in light absorption due to blood in the biological sample in relationship to time with the processor, which is a function of a standard deviation of a logarithm of the time dependent output current divided by a proportionality constant:

$$\Delta A = \frac{\sigma[A(t)]}{k} = \frac{\sigma[\log I_D(t)]}{k},$$

where  $A(t)$  is the change in light absorption due to blood in the biological sample as a function of time,  $\Delta A$  is the peak-to-valley height of  $A(t)$  plot during the time interval of one heartbeat,  $I_D(t)$  is the time dependent detector current,  $\log I_D(t)$  is the logarithm of the time dependent detector current,  $k$  is the proportionality constant  $\sigma[A(t)]$  is the standard deviation of  $A(t)$  and  $\sigma[\log I_D(t)]$  represents the standard deviation of  $\log I_D(t)$ ; and

determining a blood glucose level associated with the biological sample present in the target area based on the

16

calculated attenuance with the light absorbance change determining algorithm implemented processor.

6. A method for detecting glucose in a biological sample, comprising:

utilizing at least one light source configured to generate one or more light beams having a wavelength in a range between 800 nm and 1600 nm to strike a target area of a biological sample;

utilizing at least one photocurrent signal generating light detector positioned to receive light from the at least one light source and to generate an output photocurrent signal, having a time dependent current, which is indicative of the power of light detected;

receiving the output photocurrent signal from the at least one photocurrent signal generating light detector with a light absorbance change determining algorithm implemented processor programmed to calculate a change in a light absorption caused by blood in the biological sample based on the received output photocurrent signal;

calculating the attenuance attributable to blood in the biological sample present in the target area with a signal-to-noise ratio of at least 20-to-1;

determining a blood glucose level associated with the biological sample present in the target area based on the calculated attenuance with the light absorbance change determining algorithm implemented processor; and

wherein the step of calculating the attenuance attributable to blood in the biological sample present in the target area is based on an equation for noise reduction having at least in part on an approximation of a standard deviation of a logarithm of the time dependent output current generated by the light power from the same target area of the biological sample, wherein the equation comprises

$$\Delta A = \frac{\sigma[A(t)]}{k} = \frac{\sigma[\log I_D(t)]}{k},$$

where  $A(t)$  is the change in light absorption due to blood in the biological sample as a function of time,  $\Delta A$  is the peak-to-valley height of  $A(t)$  plot during the time interval of one heartbeat,  $I_D(t)$  is the time dependent detector current,  $\log I_D(t)$  is the logarithm of the time dependent detector current,  $k$  is the proportionality constant,  $\sigma[A(t)]$  is the standard deviation of  $A(t)$ , and  $\sigma[\log I_D(t)]$  represents the standard deviation of  $\log I_D(t)$ .

\* \* \* \* \*