



US008700115B2

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

(10) **Patent No.:** **US 8,700,115 B2**  
(45) **Date of Patent:** **\*Apr. 15, 2014**

(54) **OPTICAL SENSOR CONFIGURATION FOR RATIOMETRIC CORRECTION OF GLUCOSE MEASUREMENT**

(71) Applicant: **Glumetrics, Inc.**, Irvine, CA (US)  
(72) Inventors: **David R. Markle**, Berwyn, PA (US);  
**Soya Gamsey**, Huntington Beach, CA (US); **Thomas A. Peyser**, Menlo Park, CA (US)  
(73) Assignee: **Glumetrics, Inc.**, Irvine, CA (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 disclaimer.

(21) Appl. No.: **13/894,718**

(22) Filed: **May 15, 2013**

(65) **Prior Publication Data**  
US 2013/0267802 A1 Oct. 10, 2013

**Related U.S. Application Data**

(63) Continuation of application No. 12/612,602, filed on Nov. 4, 2009, now Pat. No. 8,467,843.

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

(52) **U.S. Cl.**  
USPC ..... **600/316; 600/309; 600/364; 600/365; 600/366**

(58) **Field of Classification Search**  
USPC ..... **600/316, 309, 317, 337, 365, 364, 366**  
See application file for complete search history.

(56) **References Cited**

U.S. PATENT DOCUMENTS

1,120,700 A	12/1914	Ehrlich
1,334,901 A	3/1920	Higdon
2,018,792 A	10/1935	Kern
2,094,224 A	9/1937	Tietz et al.
2,112,244 A	3/1938	Jurist
2,274,551 A	2/1942	Kenyon et al.
2,496,151 A	1/1950	Dawson et al.
2,812,524 A	11/1957	Pruitt

(Continued)

FOREIGN PATENT DOCUMENTS

CN	85108331	6/1987
CS	7707425	2/1980

(Continued)

OTHER PUBLICATIONS

U.S. Reexamination, Request for Inter parties Reexamination, dated Sep. 6, 2012.

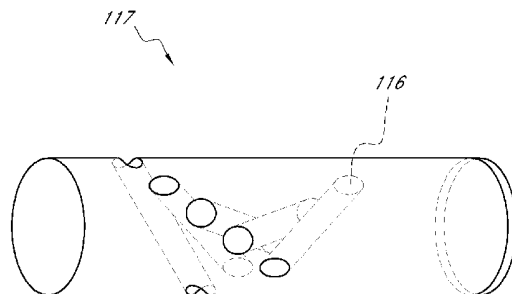
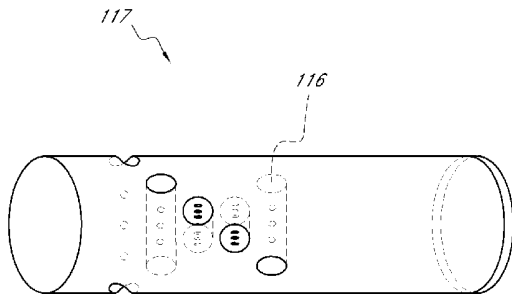
(Continued)

*Primary Examiner* — Rochelle-Ann J Blackman  
(74) *Attorney, Agent, or Firm* — Knobbe Martens Olson & Bear, LLP

(57) **ABSTRACT**

Embodiments of the invention are directed to an optical sensor for detecting glucose. The sensor comprises a chemical indicator system disposed within a gap between the distal end of an optical fiber and an atraumatic tip portion, wherein the optical fiber and atraumatic tip portion are coupled by a coupling member, such as a rod or hypotube or cage that traverses the gap. The sensor further comprises a means for generating and detecting an optical reference signal unrelated to the glucose, such that ratiometric correction of glucose measurements for artifacts in the optical system is enabled.

**31 Claims, 20 Drawing Sheets**



(56)

## References Cited

## U.S. PATENT DOCUMENTS

3,011,293 A	12/1961	Rado	4,796,633 A	1/1989	Zwirkoski
3,302,219 A	2/1967	Harris	4,798,738 A	1/1989	Yafuso et al.
3,488,098 A	1/1970	Sobczak	4,801,187 A	1/1989	Elbert et al.
3,659,586 A	5/1972	Johns et al.	4,803,049 A	2/1989	Hirschfeld et al.
3,795,239 A	3/1974	Eberhard et al.	4,810,655 A	3/1989	Khalil et al.
3,827,089 A	8/1974	Grow	4,816,130 A	3/1989	Karakelle et al.
3,846,353 A	11/1974	Grotta	4,820,636 A	4/1989	Hill et al.
3,865,548 A	2/1975	Padawer	4,821,738 A	4/1989	Iwasaki et al.
3,874,010 A	4/1975	Geary	4,822,127 A	4/1989	Kamiya et al.
3,884,225 A	5/1975	Witter	4,833,091 A	5/1989	Leader et al.
3,895,403 A	7/1975	Davis	4,838,269 A	6/1989	Robinson et al.
3,905,888 A	9/1975	Mindt et al.	4,844,841 A	7/1989	Koller et al.
3,909,504 A	9/1975	Browne	4,846,543 A	7/1989	Kapany et al.
3,924,281 A	12/1975	Gibbs	4,851,195 A	7/1989	Matthews et al.
3,930,580 A	1/1976	Bazell et al.	4,854,321 A	8/1989	Boiarski
3,996,345 A	12/1976	Ullman	4,861,728 A	8/1989	Wagner
4,003,707 A	1/1977	Lübbbers et al.	4,872,226 A	10/1989	Lonardo
4,041,932 A	8/1977	Fostick	4,872,759 A	10/1989	Stich-Baumeister
4,094,578 A	6/1978	DiVita et al.	4,886,338 A	12/1989	Yafuso et al.
4,180,879 A	1/1980	Mann	4,889,407 A	12/1989	Markle et al.
4,197,853 A	4/1980	Parker	4,903,707 A	2/1990	Knute et al.
4,200,110 A	4/1980	Peterson et al.	4,906,232 A	3/1990	Reynolds
4,240,438 A	12/1980	Updike et al.	4,923,273 A	5/1990	Taylor
4,245,645 A	1/1981	Arseneault et al.	4,927,222 A	5/1990	Kamiya et al.
4,269,605 A	5/1981	Dean et al.	4,937,901 A	7/1990	Brennan
4,306,562 A	12/1981	Osborne	4,939,801 A	7/1990	Schaal et al.
4,307,933 A	12/1981	Palmer et al.	4,941,308 A	7/1990	Grabenkort et al.
4,308,254 A	12/1981	Tayot et al.	4,943,364 A	7/1990	Koch et al.
4,344,438 A	8/1982	Schultz	4,946,038 A	8/1990	Eaton
4,345,606 A	8/1982	Littleford	4,955,862 A	9/1990	Sepetka
4,358,851 A	11/1982	Scilfres et al.	4,960,412 A	10/1990	Fink
4,361,918 A	12/1982	Roisseth	4,966,597 A	10/1990	Cosman
4,371,374 A	2/1983	Cerami et al.	5,000,901 A	3/1991	Iyer et al.
4,459,712 A	7/1984	Pathan	5,005,576 A	4/1991	Günther
4,465,335 A	8/1984	Eppes	5,007,704 A	4/1991	McCartney
4,469,357 A	9/1984	Martin	5,012,809 A	5/1991	Shulze
4,474,431 A	10/1984	Bricheno	5,018,225 A	5/1991	Fergni et al.
4,476,870 A	10/1984	Peterson et al.	5,030,420 A	7/1991	Bacon
4,490,867 A	1/1985	Gabrielson	5,047,020 A	9/1991	Hsu
4,495,293 A	1/1985	Shaffar	5,047,208 A	9/1991	Schweitzer et al.
4,502,169 A	3/1985	Persson	5,047,627 A	9/1991	Yim et al.
RE31,879 E	5/1985	Lübbbers et al.	5,054,497 A	10/1991	Kapp et al.
4,528,616 A	7/1985	Koppensteiner	5,068,931 A	12/1991	Smith
4,548,907 A	10/1985	Seitz et al.	5,069,674 A	12/1991	Fearnot et al.
4,557,900 A	12/1985	Heitzmann	5,082,112 A	1/1992	Dunklee
4,560,248 A	12/1985	Cramp et al.	5,093,266 A	3/1992	Leader et al.
4,568,444 A	2/1986	Nakamura et al.	5,098,618 A	3/1992	Zelez
4,600,310 A	7/1986	Cramp et al.	5,104,388 A	4/1992	Quackenbush
4,621,049 A	11/1986	Wang	5,108,502 A	4/1992	Pawlowski et al.
4,629,451 A	12/1986	Winters et al.	5,109,452 A	4/1992	Selvin et al.
4,636,144 A	1/1987	Abe et al.	5,114,676 A	5/1992	Leiner et al.
4,649,271 A	3/1987	Hök et al.	5,119,463 A	6/1992	Vurek
4,650,472 A	3/1987	Bates	5,132,432 A	7/1992	Haugland et al.
4,654,031 A	3/1987	Lentz	5,137,033 A	8/1992	Norton
4,654,127 A	3/1987	Baker et al.	5,137,833 A	8/1992	Russell
4,654,327 A	3/1987	Teng	5,141,497 A	8/1992	Erskine
4,659,817 A	4/1987	Gallop et al.	5,156,962 A	10/1992	Suzuki et al.
4,675,925 A	6/1987	Littleton	5,162,130 A	11/1992	McLaughlin
4,684,538 A	8/1987	Klemarczyk	5,166,990 A	11/1992	Riccitelli et al.
4,689,308 A	8/1987	Gerhard	5,167,715 A	12/1992	Kalafala et al.
RE32,514 E	10/1987	Steklenski	5,168,587 A	12/1992	Shutes
4,703,756 A	11/1987	Gough et al.	5,175,016 A	12/1992	Yafuso et al.
4,707,056 A	11/1987	Bittner	5,176,882 A	1/1993	Gray et al.
4,710,623 A	12/1987	Lipson et al.	5,178,267 A	1/1993	Grabenkort et al.
4,727,730 A	3/1988	Boiarski et al.	5,180,376 A	1/1993	Fischell
4,737,153 A	4/1988	Shimamura et al.	5,182,353 A	1/1993	Hui et al.
4,744,618 A	5/1988	Mahlein	5,185,263 A	2/1993	Kroneis et al.
4,746,751 A	5/1988	Oviatt	5,188,803 A	2/1993	Hochberg
4,750,795 A	6/1988	Blotekjaer	5,217,691 A	6/1993	Greene et al.
4,751,918 A	6/1988	Bernard et al.	5,230,031 A	7/1993	Markle
4,754,538 A	7/1988	Stewart, Jr. et al.	5,246,109 A	9/1993	Markle et al.
4,776,047 A	10/1988	DiMatteo	5,246,441 A	9/1993	Ross et al.
4,785,814 A	11/1988	Kane	5,257,338 A	10/1993	Markle
4,792,689 A	12/1988	Peterson	5,262,037 A	11/1993	Markle et al.
4,794,619 A	12/1988	Tregay	5,279,596 A	1/1994	Castaneda et al.
			5,280,130 A	1/1994	Markle et al.
			5,280,548 A	1/1994	Atwater et al.
			5,284,154 A	2/1994	Raymond et al.
			5,286,294 A	2/1994	Ebi et al.

(56)

## References Cited

## U.S. PATENT DOCUMENTS

5,290,266	A	3/1994	Rohling et al.	6,585,665	B1	7/2003	Chapman et al.
5,302,731	A	4/1994	Pitner et al.	6,602,702	B1	8/2003	McDevitt et al.
5,305,740	A	4/1994	Kolobow	6,623,490	B1	9/2003	Crane et al.
5,310,471	A	5/1994	Markle et al.	6,627,177	B2	9/2003	Singaram et al.
5,312,344	A	5/1994	Grinfeld	6,653,141	B2	11/2003	Singaram et al.
5,322,513	A	6/1994	Walker	6,663,595	B2	12/2003	Spohn et al.
5,330,718	A	7/1994	Hui et al.	6,682,938	B1	1/2004	Satcher, Jr. et al.
5,334,157	A	8/1994	Klein et al.	6,702,972	B1	3/2004	Markle
5,335,305	A	8/1994	Kosa et al.	6,711,423	B2	3/2004	Colvin
5,354,448	A	10/1994	Markle et al.	6,741,877	B1	5/2004	Shults et al.
5,357,732	A	10/1994	Markle et al.	6,766,183	B2	7/2004	Walsh et al.
5,361,758	A	11/1994	Hall et al.	6,794,195	B2	9/2004	Colvin
5,380,304	A	1/1995	Parker	6,800,451	B2	10/2004	Daniloff et al.
5,389,217	A	2/1995	Singer	6,804,544	B2	10/2004	Van Antwerp et al.
5,408,999	A	4/1995	Singh et al.	6,855,556	B2	2/2005	Amiss et al.
5,409,469	A	4/1995	Schaerf	6,858,403	B2	2/2005	Han et al.
5,503,770	A	4/1996	James et al.	7,033,322	B2	4/2006	Silver
5,511,408	A	4/1996	Yoshioka et al.	7,064,103	B2	6/2006	Pitner et al.
5,511,547	A	4/1996	Markle et al.	D525,632	S	7/2006	Jost et al.
5,512,246	A	4/1996	Russell et al.	RE39,438	E	12/2006	Shah et al.
5,514,710	A	5/1996	Haugland et al.	7,181,260	B2	2/2007	Gutierrez
5,536,783	A	7/1996	Olstein et al.	7,216,001	B2	5/2007	Hacker et al.
5,545,179	A	8/1996	Williamson, IV	D544,871	S	6/2007	Lim et al.
5,558,714	A	9/1996	Watanabe et al.	7,226,414	B2	6/2007	Ballerstadt
5,569,186	A	10/1996	Lord et al.	7,229,450	B1	6/2007	Chitre et al.
5,596,988	A	1/1997	Markle et al.	D550,242	S	9/2007	Nijijima
5,605,152	A	2/1997	Slate et al.	D550,245	S	9/2007	Nijijima
5,618,587	A	4/1997	Markle et al.	7,276,029	B2	10/2007	Goode, Jr. et al.
5,622,259	A	4/1997	Church	7,277,740	B2	10/2007	Rohleder et al.
5,634,911	A	6/1997	Hermann et al.	7,277,745	B2	10/2007	Natarajan et al.
5,643,212	A	7/1997	Coutre et al.	7,303,814	B2	12/2007	Lamberti et al.
5,643,580	A	7/1997	Subramaniam	7,306,621	B1	12/2007	Halla et al.
5,658,264	A	8/1997	Samson	D559,264	S	1/2008	Nijijima
5,669,920	A	9/1997	Conley et al.	D560,224	S	1/2008	Park et al.
5,676,784	A	10/1997	McGaffigan	7,316,909	B2	1/2008	Pitner et al.
D388,418	S	12/1997	Polson et al.	7,317,111	B2	1/2008	Bhatt et al.
5,700,253	A	12/1997	Parker	D610,065	S	2/2008	Part et al.
5,702,373	A	12/1997	Samson	7,326,538	B2	2/2008	Pitner et al.
5,747,666	A	5/1998	Willis	7,345,160	B2	3/2008	Daunert et al.
5,755,704	A	5/1998	Lunn	7,353,055	B2	4/2008	Hogan
5,763,238	A	6/1998	James et al.	7,358,094	B2	4/2008	Bell et al.
5,797,876	A	8/1998	Spears et al.	7,381,184	B2	6/2008	Funderburk et al.
5,810,985	A	9/1998	Bao et al.	7,381,938	B2	6/2008	Kobayashi et al.
5,827,242	A	10/1998	Follmer et al.	7,390,462	B2	6/2008	Rao et al.
5,891,100	A	4/1999	Fleckenstein	7,417,164	B2	8/2008	Suri
5,891,114	A	4/1999	Chien et al.	D580,950	S	11/2008	Steele et al.
5,922,612	A	7/1999	Alder et al.	D582,939	S	12/2008	Neuhaus
5,947,940	A	9/1999	Beisel	7,470,420	B2	12/2008	Singaram et al.
5,951,929	A	9/1999	Wilson	7,496,392	B2	2/2009	Alarcon et al.
5,954,651	A	9/1999	Berg et al.	D592,223	S	5/2009	Neuhaus
6,002,954	A	12/1999	Van Antwerp et al.	7,559,894	B2	7/2009	McEowen
6,011,984	A	1/2000	Van Antwerp et al.	7,615,007	B2	11/2009	Shults et al.
6,019,736	A	2/2000	Avellanet et al.	7,661,301	B2	2/2010	Moor
6,117,290	A	9/2000	Say et al.	7,751,863	B2	7/2010	Markle et al.
6,152,933	A	11/2000	Werp et al.	7,767,846	B2	8/2010	Suri
6,156,010	A	12/2000	Kuracina et al.	D626,143	S	10/2010	Karten et al.
6,175,752	B1	1/2001	Say et al.	7,824,918	B2	11/2010	Suri
6,187,130	B1	2/2001	Berard et al.	7,829,341	B2	11/2010	Gamsey et al.
6,197,534	B1	3/2001	Lakowicz et al.	7,879,024	B2	2/2011	Thorstenson et al.
6,200,301	B1	3/2001	Pfeiffer et al.	7,881,780	B2	2/2011	Flaherty
6,227,627	B1	5/2001	Goossens	7,939,664	B2	5/2011	Gamsey et al.
6,254,829	B1	7/2001	Hartmann et al.	7,959,577	B2	6/2011	Schmitz et al.
6,273,874	B1	8/2001	Parris	7,981,058	B2	7/2011	Akay
6,304,766	B1	10/2001	Colvin, Jr.	8,088,097	B2	1/2012	Markle et al.
6,319,540	B1	11/2001	Van Antwerp et al.	8,110,251	B2	2/2012	Markle et al.
6,361,508	B1	3/2002	Johnson et al.	8,178,676	B2	5/2012	Gamsey et al.
6,363,273	B1	3/2002	Mastrorio et al.	8,202,731	B2	6/2012	Suri
6,370,406	B1	4/2002	Wach et al.	8,467,843	B2*	6/2013	Markle et al. .... 600/316
6,375,627	B1	4/2002	Mauze	8,473,222	B2	6/2013	Romey et al.
6,387,672	B1	5/2002	Arimori et al.	2001/0016682	A1	8/2001	Berner et al.
6,464,849	B1	10/2002	Say et al.	2002/0018843	A1	2/2002	Van Antwerp et al.
6,477,395	B2	11/2002	Schulman et al.	2002/0026108	A1	2/2002	Colvin, Jr.
6,521,447	B2	2/2003	Zou et al.	2002/0107178	A1	8/2002	Van Den Berghe
6,544,212	B2	4/2003	Galley et al.	2002/0128546	A1	9/2002	Silver
6,584,335	B1	6/2003	Haar et al.	2002/0193672	A1	12/2002	Walsh et al.
				2003/0013974	A1	1/2003	Natarajan et al.
				2003/0028089	A1	2/2003	Galley et al.
				2003/0065254	A1	4/2003	Schulman et al.
				2003/0100821	A1	5/2003	Heller et al.

(56)

## References Cited

## U.S. PATENT DOCUMENTS

2003/0171666 A1 9/2003 Loeb  
 2003/0232198 A1 12/2003 Lamberti et al.  
 2003/0232383 A1 12/2003 Daunert et al.  
 2004/0028612 A1 2/2004 Singaram et al.  
 2004/0072358 A1 4/2004 Ballerstadt  
 2004/0077969 A1 4/2004 Onda et al.  
 2004/0087842 A1 5/2004 Lakowicz et al.  
 2004/0136924 A1 7/2004 Boyd et al.  
 2004/0176672 A1 9/2004 Silver et al.  
 2004/0219535 A1 11/2004 Bell et al.  
 2004/0230138 A1 11/2004 Inoue et al.  
 2004/0243018 A1 12/2004 Organ et al.  
 2004/0260158 A1 12/2004 Hogan  
 2004/0260162 A1 12/2004 Rohleder et al.  
 2004/0267203 A1 12/2004 Potter et al.  
 2005/0019219 A1 1/2005 Oshiman et al.  
 2005/0048121 A1 3/2005 East et al.  
 2005/0054975 A1 3/2005 Patel et al.  
 2005/0059097 A1 3/2005 Daunert et al.  
 2005/0090014 A1 4/2005 Rao et al.  
 2005/0113658 A1 5/2005 Jacobson et al.  
 2005/0118599 A1 6/2005 Pawliszyn  
 2005/0118726 A1 6/2005 Schultz et al.  
 2005/0123935 A1 6/2005 Haugland et al.  
 2005/0124896 A1 6/2005 Richter et al.  
 2005/0130249 A1 6/2005 Parris et al.  
 2005/0173245 A1 8/2005 Feldman et al.  
 2005/0193860 A1 9/2005 Schulman et al.  
 2005/0233465 A1 10/2005 Miller  
 2005/0240086 A1 10/2005 Akay  
 2005/0241959 A1 11/2005 Ward et al.  
 2005/0266038 A1 12/2005 Glauser et al.  
 2005/0267326 A1 12/2005 Loeb et al.  
 2005/0282225 A1 12/2005 Daunert et al.  
 2005/0283204 A1 12/2005 Buhlmann et al.  
 2006/0008500 A1 1/2006 Chavan et al.  
 2006/0039950 A1 2/2006 Zhou et al.  
 2006/0051874 A1 3/2006 Reed et al.  
 2006/0079740 A1 4/2006 Silver et al.  
 2006/0083688 A1 4/2006 Singaram et al.  
 2006/0084854 A1 4/2006 Cho et al.  
 2006/0088722 A1 4/2006 Aller et al.  
 2006/0105174 A1 5/2006 Aller et al.  
 2006/0135888 A1 6/2006 Mimmagh-Kelleher et al.  
 2006/0173252 A1 8/2006 Ellingsen et al.  
 2006/0189863 A1 8/2006 Peyser et al.  
 2006/0189979 A1 8/2006 Esch et al.  
 2006/0195042 A1 8/2006 Flahert  
 2006/0270949 A1 11/2006 Mathie et al.  
 2006/0287600 A1 12/2006 McEowen  
 2007/0003588 A1 1/2007 Chinn et al.  
 2007/0014726 A1 1/2007 Merial et al.  
 2007/0020181 A1 1/2007 Workman et al.  
 2007/0038155 A1 2/2007 Kelly, Jr. et al.  
 2007/0060872 A1 3/2007 Hall et al.  
 2007/0100356 A1 5/2007 Lucatero et al.  
 2007/0123775 A1 5/2007 Meyer et al.  
 2007/0136825 A1 6/2007 Frommer et al.  
 2007/0156079 A1 7/2007 Brown  
 2007/0175828 A1 8/2007 Goedje et al.  
 2007/0179437 A1 8/2007 Grage et al.  
 2007/0207498 A1 9/2007 Palmieri et al.  
 2007/0244382 A1 10/2007 Robinson et al.  
 2007/0256477 A1 11/2007 Moor  
 2007/0265675 A1 11/2007 Lund et al.  
 2008/0001091 A1 1/2008 Kobayashi et al.  
 2008/0009687 A1 1/2008 Smith et al.  
 2008/0027245 A1 1/2008 Suri  
 2008/0086042 A1 4/2008 Brister et al.  
 2008/0154101 A1 6/2008 Jain et al.  
 2008/0154107 A1 6/2008 Jina  
 2008/0183061 A1 7/2008 Goode et al.  
 2008/0188722 A1 8/2008 Markle et al.  
 2008/0188725 A1 8/2008 Markle et al.  
 2008/0261255 A1 10/2008 Tolosa et al.

2008/0287761 A1 11/2008 Hayter et al.  
 2008/0305009 A1 12/2008 Gamsey et al.  
 2008/0305506 A1 12/2008 Suri  
 2008/0306363 A1 12/2008 Chaiken et al.  
 2009/0018418 A1 1/2009 Markle et al.  
 2009/0018426 A1 1/2009 Markle et al.  
 2009/0048430 A1 2/2009 Hellinga et al.  
 2009/0061528 A1 3/2009 Suri  
 2009/0062696 A1 3/2009 Nathan et al.  
 2009/0076336 A1 3/2009 Mazar et al.  
 2009/0081803 A1 3/2009 Gamsey et al.  
 2009/0082566 A1 3/2009 Mitra  
 2009/0088329 A1 4/2009 Brennan et al.  
 2009/0104714 A1 4/2009 Thomas et al.  
 2009/0112075 A1 4/2009 Klok et al.  
 2009/0143659 A1 6/2009 Li et al.  
 2009/0171381 A1 7/2009 Schmitz et al.  
 2009/0177143 A1 7/2009 Markle et al.  
 2009/0182217 A1 7/2009 Li et al.  
 2009/0192416 A1 7/2009 Ernst et al.  
 2009/0196864 A1 8/2009 Bulla  
 2009/0200620 A1 8/2009 Omura et al.  
 2009/0228068 A1 9/2009 Buhlmann et al.  
 2009/0264719 A1 10/2009 Markle et al.  
 2009/0324945 A1 12/2009 Licht et al.  
 2010/0173065 A1 7/2010 Michal et al.  
 2010/0274110 A1 10/2010 Markle  
 2010/0279424 A1 11/2010 Suri  
 2010/0292617 A1 11/2010 Lei et al.  
 2010/0312483 A1 12/2010 Peysr  
 2011/0077477 A1 3/2011 Romey et al.  
 2011/0105866 A1 5/2011 Markle  
 2011/0152658 A1 6/2011 Peyser  
 2011/0171742 A1 7/2011 Gamsey  
 2011/0224516 A1 9/2011 Romey et al.  
 2011/0236989 A1 9/2011 Suri et al.  
 2011/0263953 A1 10/2011 Markle  
 2012/0053427 A1 3/2012 Markle  
 2012/0116191 A1 5/2012 Markle  
 2012/0208286 A1 8/2012 Gamsey et al.  
 2012/0282412 A1 11/2012 Markle

## FOREIGN PATENT DOCUMENTS

DE 3036868 5/1982  
 DE 3509262 10/1985  
 DE 3720736 1/1989  
 DE 195 02 183 8/1996  
 DE 298 17 986 2/1999  
 DE 198 20 808 11/1999  
 EP 0 073 558 3/1983  
 EP 0 147 168 7/1985  
 EP 0 596 700 5/1994  
 EP 0 617 977 A1 10/1994  
 EP 0 758 451 B1 1/1999  
 EP 000760723-0001 7/2007  
 EP 2 217 316 7/2010  
 EP 2 222 686 8/2010  
 EP 2 147 003 4/2011  
 EP 2 054 476 6/2011  
 FR 2 350 831 12/1977  
 FR 2 624 007 6/1989  
 GB 1 123 094 8/1968  
 GB 1 447 163 8/1976  
 GB 2 048 682 12/1980  
 JP 53-68249 6/1978  
 JP 54-13347 1/1979  
 JP 54-111363 8/1979  
 JP 54-155856 12/1979  
 JP 56-116752 9/1981  
 JP 56-116754 9/1981  
 JP 58-162921 9/1983  
 JP 3-52936 3/1991  
 JP 06-016859 4/1994  
 JP 06-285049 10/1994  
 JP 2003-262613 9/2003  
 JP 1332866 5/2008  
 JP 2009-544729 12/2009  
 JP 2010-507711 3/2010

(56)

## References Cited

## FOREIGN PATENT DOCUMENTS

JP	2010-517693	5/2010
JP	2010-518397	5/2010
JP	2010-526599	8/2010
JP	2010-527010	8/2010
JP	2010-535903	11/2010
JP	2011-504399	2/2011
JP	2011-511755	4/2011
JP	5017377	6/2012
SU	6216724	8/1978
WO	WO 87/00920	2/1987
WO	WO 88/04415	6/1988
WO	WO 92/19150	11/1992
WO	WO 94/10553	5/1994
WO	WO 96/22730	8/1996
WO	WO 96/22798	8/1996
WO	WO 97/20530	6/1997
WO	WO 97/37713	10/1997
WO	WO 97/48437	12/1997
WO	WO 98/08554	3/1998
WO	WO 00/02048	1/2000
WO	WO 00/13003	3/2000
WO	WO 00/43536	7/2000
WO	WO 01/20019	3/2001
WO	WO 02/46752	6/2002
WO	WO 03/034047	4/2003
WO	WO 03/060464	7/2003
WO	WO 2004/054438	7/2004
WO	WO 2004/099778	11/2004
WO	WO 2005/090014	4/2005
WO	WO 2005/054831	6/2005
WO	WO 2005/065241	7/2005
WO	WO 2006/023725	3/2006
WO	WO 2006/044973	4/2006
WO	WO 2007/059311	5/2007
WO	WO 2007/067743	6/2007
WO	WO 2007/105140	9/2007
WO	WO 2008/001091	1/2008
WO	WO 2008/014280	1/2008
WO	WO 2008/072338	6/2008
WO	WO 2008/097747	8/2008
WO	WO 2008/098011	8/2008
WO	WO 2008/098087	8/2008
WO	WO 2008/137604	11/2008
WO	WO 2008/141241	11/2008
WO	WO 2008/141243	11/2008
WO	WO 2009/009756	1/2009
WO	WO 2009/018426	1/2009
WO	WO 2009/021057	2/2009
WO	WO 2009/067626	5/2009
WO	WO 2009/129186	10/2009
WO	WO 2010/123972	10/2010
WO	WO 2010/141888	12/2010
WO	WO 2011/041546	4/2011
WO	WO 2011/056274	5/2011
WO	WO 2011/075710	6/2011
WO	WO 2011/075711	6/2011
WO	WO 2011/084713	7/2011
WO	WO 2011/097586	8/2011
WO	WO 2011/112020	9/2011
WO	WO 2011/137178	11/2011
WO	WO 2013/033076	3/2013
WO	WO 2013/049068	4/2013

## OTHER PUBLICATIONS

U.S. Transmittal of Communication to Third Party Requester Inter Partes Reexamination dated Oct. 22, 2012.

Atherton, S. J., et al.: "Reactions of Three Bis(viologen) Tetraquaternary Salts and Their Reduced Radicals", *J. Am. Chem. Soc.* 1986, 108,3380-3385.

Ayala et al., Database Caplus, DN 133:189758. (*Protein Science* (2000), 9(8), 1589-1593).

Badugu, R., et al.: "A Glucose sensing contact lens: A new approach to non-invasive continuous physiological glucose monitoring", *SPIE*

Proceedings, The International Society for Optical Engineering—SPIE, Bellingham, Washington, USA, vol. 5317, Jan. 25, 2004.

Baldini "Invasive Sensors in Medicine." *Optical Chemical Sensors*, NATO Science Series 11: Mathematics, Physics and Chemistry [online], 2006 [Retrieved on Nov. 15, 2010], vol. 224, pp. 417-435, Retrieved from the Internet: <URL <http://www.springerlink.com>>.

Ballerstadt, Ralph, et al.: "Fluorescence Resonance Energy Transfer-Based Near-Infrared Fluorescence Sensor for Glucose Monitoring", *Diabetes Technology & Therapeutics*, vol. 6, No. 2, Apr. 1, 2004.

Benmakroha et al. "Haemocompatibility of invasive sensors," *Med. & Biol. Eng. & Comput.*, 1995, 33,811-821 (Nov. 1995).

Bolton C F. 1999 Acute Weakness. In: *Oxford Textbook of Critical Care*; Eds. Webb A R, Shapiro M J, Singer M, Suter P M; Oxford Medical Publications, Oxford UK; pp. 490-495.

Cao, H., et al.: "Fluorescent Chemosensors for Carbohydrates: A Decade's Worth of Bright Spies for Saccharides in Review", *Journal of Fluorescence*, vol. 14, No. 5, Sep. 2004.

Check, W., "ICUs tighten belts on blood glucose levels", *Cap Today*, Feb. 2005, in 7 pages, vol. 19-2, pp. 1,95-96,98,100,102, and 104 ("Check").

Choleau et al.: "Calibration of a subcutaneous amperometric glucose sensor implanted for 7 days in diabetic patients. Part 2. Superiority of the one-point calibration method." *Biosensors and Bioelectronics*, vol. 17, No. 8, Aug. 1, 2002.

Cordes, D. B., et al.: "The Interaction of Boronic Acid-Substituted Viologens with Pyranine: The Effects of Quencher Charge on Fluorescence Quenching and Glucose Response", *Langmuir* 2005,21, 6540-6547.

DiCesare, N., et al.: "Saccharide Detection Based on the Amplified Fluorescence Quenching of a Water-Soluble Poly(phenylene ethynylene) by a Boronic Acid Functionalized Benzyl Viologen Derivative", *Langmuir* 2002, 18, 7785-7787.

"Fiber Optic Oxygen Sensors: Theory of Operation", *Fiber Optic Oxygen Sensors Theory of Operation*, <http://www.oceanoptics.com/Products/sensorththeory.asp>—4 pages.

Fidaleo et al., Database Caplus, DN 140:249134 (*Chemical and Biochemical Engineering Quarterly* (2003), 17(4), 311-318).

Furnary A.P. et al. "Effect of hyperglycemia and continuous intravenous insulin infusions on outcomes of cardiac surgical procedures: The Portland Diabetic Project", *Endocrine Practice*, Mar./Apr. 2004, pp. 21-33, vol. 10.

Hirsch Irl B. et al. "Acute Complications of Diabetes" *Endocrinology and Metabolism Clinics of North America*, Dec. 2000, pp. 745-771, vol. 29-4.

Hunneche, "Antioxidant Activity of a Combinatorial Library of Emulsifier—Antioxidant Bioconjugates," *J. Agric. Food Chem.* 2008, 56, 9258-9268.

Hvastkovs, E. G., et al.: "Minor Groove Binding of a Novel Tetracationic Diviologen", *Langmuir* 2006, 22, 10821-10829.

Kuraganov, B. I., et al.: Criterion for Hill equation validity for description of biosensor calibration curves, *Analytica Chimica Acta*, vol. 427, No. 1, Jan. 1, 2001.

Kuwabara, T., et al.: "Effect of Alkali Metal Ions on Photochromic Behavior of Bisviologen-incorporated Oligo-oxethylene Units", *Rapid Communication. Photochemistry and Photobiology*, 2003, 77(5); 572-575.

Lee, S. K., et al.: "Conformation and binding properties of polymethylene-linked bisviologens-2-naphthol complexes", *Journal of the Chemical Society, Perkin Transactions 2* 2001, 1983-1988.

Leijten FSS & De Weerd A W 1994 Critical illness polyneuropathy: a review of the literature, definition and pathophysiology. *Clinical Neurology and Neurosurgery*, 96: 10-19.

Levetan et al., "Hospital Management of Diabetes," in *Acute Complications of Diabetes*, vol. 29, No. 4, 745-71, at 745-54 (Dec. 2000).

Liu, et al., "Characterization of Immobilization of an Enzyme in a Modified Y Zeolite Matrix and Its Application to an Amperometric Glucose Biosensor," *Anal. Chem.* 1997, 69, pp. 2343-2348.

Mignani et al. "Biomedical sensors using optical fibres." *Reports on Progress in Physics* [online], Jan. 1996 [Retrieved on Nov. 15, 2010], vol. 59, No. 1, pp. 1-28, Retrieved from the internet: <URL <http://iopscience.iop.org>>.

Mizock B A. *Am J Med* 1995; 98: 75-84.

(56)

## References Cited

## OTHER PUBLICATIONS

- Mohr, G. J. et al.: Application of a Novel Lipophilized Fluorescent dye in an Optical Nitrate Sensor, *Journal of Fluorescence* 1995, 5, 135-138.
- "Optical Glucose Sensor Holds Promise for Diabetics and Intensive Care Patients," in *Science Daily*, Mar. 18, 2004 (archived on Apr. 4, 2004 at: <<http://web.archive.org/web/20040404161607/http://www.ScienceDaily.com/releases/2004/03/0403529.htm>> ("ScienceDaily Article").
- Park, Y. S., et al.: "Facile Reduction of Zeolite Encapsulated Viologens with Solvated Electrons and Selective Dispersion of Inter- and Intramolecular Dimers of Propylene-Bridged Bisviologen Radical Cation", *Langmuir* 2000, 16, 4470-4477.
- PCT International Report on Patentability re PCT/US2010/044761, dated May 9, 2012.
- Peterson et al. "Fiber-optic for in vivo measurement of oxygen partial pressure," *Analytical Chemistry* [online], Jan. 1984 [Retrieved on Nov. 15, 2010], vol. 57, No. 1, pp. 62-67, Retrieved from the Internet: <URL: <http://pubs.acs.org>>.
- Piper, Hannah G. "Real-Time Continuous Glucose Monitoring in Pediatric Patients During and After Cardiac Surgery," in *Pediatrics*, vol. 118, No. 3, Sep. 2006.
- Retrieved from the Internet <URL: [Journal of Agricultural and Food Chemistry, vol. 53, No. 23, Nov. 1, 2005.](http://www.invitrogen.com/site/us/en/home/References/Molecular-Probes-The-Handbook/Fluorophores-and-Their-Amine-Reactive-Derivatives/Fluorescein-Oregon-Green-and-Rhodamine-Green-Dyes.htm1#></a>, 11 pages.</p>
<p>Reyes-De-Corcuera, Josi et al.: )
- Sato, H., et al.: "Polymer Effect in Electrochromic Behavior of Oligomeric Viologens", *Journal of Applied Polymer Science*, vol. 24, 2075-2085 (1979).
- Stokes, et al.: "An optical oxygen sensor and reaction vessel for high-pressure applications", *Limnol. Oceanogr.*, 44(1), 1999, 189-195.
- Su et al., "Polyethersulfone Hollow Fiber Membranes for Hemodialysis," *Progress in Hemodialysis—From Emergent Biotechnology to Clinical Practice*, [www.intechopen.com](http://www.intechopen.com), Nov. 7, 2011. ISBN 978-953-307-377-4.
- Suri, J. T. et al.: "Monosaccharide Detection with 4,7-Phenanthroline Salts: Charge-Induced Fluorescence Sensing", *Langmuir* 2003, 19, 5145-5152.
- Takashima, H., et al.: "Remarkably stereoselective photoinduced electron-transfer reaction between zinc myoglobin and optically active binaphthyl bisviologen", *Journal of Biological Inorganic Chemistry* 2003, 8, 499-506.
- Tsukahara, K., et al.: "Syntheses, Characterizations, and Redox Behavior of Optically Active Viologens and Bisviologens", *Bulletin of the Chemical Society of Japan* 1999, 72, 139-149.
- Van Den Berghe G., et al. "Intensive Insulin Therapy in the Medical ICU", *The New England Journal of Medicine*, Feb. 2, 2006, pp. 449-461, vol. 354-5.
- Volker, Ludwig et al.: "Continuous Glucose Monitoring with Glucose Sensors: Calibration and Assessment Criteria", *Diabetes Technology & Therapeutics*, vol. 5, No. 4, Aug. 1, 2003.
- Wilson, Intensive Insulin Therapy in Critical Care, *Diabetes Care*, Apr. 2007, pp. 1005-1011, vol. 30-4.
- Zisser, MD, et al. "Excitation: The use of Fluorescence in Glucose Monitoring (Part 11)," *Glumetrics*, Feb. 13, 2010.
- Zochodne D W et al. 1987 Polyneuropathy associated with critical illness: a complication of sepsis and multiple organ failure. *Brain*, 110: 819-842.
- Agayn, V. I. and Dr. R. Walt (1993). "Fiber-optic sensor for continuous monitoring of fermentation pH." *Biotechnology* 72(6):6-9.
- Angel, S. M., "Optrodes: Chemically Selective Fiber Optic Sensors," *Spectroscopy*, Apr. 1987, pp. 38-47.
- Badugu R., et al., "Wavelength-ratiometric near-physiological pH sensors based on 6-aminoquinolinium boronic acid probes" *Talanta*, Elsevier, Amsterdam, NL, Apr. 30, 2005, vol. 66, Issue No. 3, pp. 569-574.
- Badugu, R. et al. "Boronic acid fluorescent sensors for monosaccharide signaling based on the 6-methoxyquinolinium heterocyclic nucleus: progress toward noninvasive and continuous glucose monitoring." *2005 Bioorg. Med. Chem.* 13 (1):113-119.
- Badugu, R. et al. "Fluorescence sensors for monosaccharides based on the 6-methylquinolinium nucleus and boronic acid moiety: potential application to ophthalmic diagnostics." *2005 Talanta* 65 (3):762-768.
- Bean & Johnson, 54 *J. Am. Chem. Soc.* 4415 (1932).
- Burnett, Peebles & Hageman, 96 *Biochemical and Biophysical Research Communications* 157 (1980).
- Cappuccio, F.E. et al. 2004 "Evaluation of pyranine derivatives in boronic acid based saccharide sensing: Significance of charge interaction between dye and quencher in solution and hydrogel" *Journal of Fluorescence* 14:521-533.
- Cordes, D. B., et al., 2006, in *Topics in Fluorescence Spectroscopy*; vol. 11, *Glucose Sensing*, Springer "Two component optical sugar sensing using boronic acid-substituted viologens with anionic fluorescent dyes" pp. 47-87. (ISBN: 978-0-387-29571-8 p. 76, scheme 3.7).
- Dawson, et al., 98 *JACS* 5996 (1970).
- Definition of "catheter" from Webster's Ninth New Collegiate Dictionary, 1990, p. 216.
- EPO Exam Report re EP App. No. 08 728 399.0, dated Dec. 7, 2010.
- European Examination Report dated May 11, 2010, re EP Application No. 08 729 209.0.
- European Examination Report dated Jan. 25, 2012, re EP Application No. 08 729 209.0.
- European Examination Report dated Jul. 2, 2012, re EP Application No. 08 729 209.0.
- EPO Office Action re App. No. 07 799 791.4 dated Jan. 29, 2010.
- European Examination Report re Application No. 08 755 267.5, dated Apr. 26, 2010.
- European Examination Report re Application No. 08 755 267.5, dated Sep. 14, 2010.
- European Examination Report re App. No. 08 797 302.0, dated Nov. 7, 2011.
- European Examination Report re App. No. 08 797 302.0, dated Jan. 24, 2011.
- European Examination dated Apr. 1, 2010, re EP Application No. 08 769 266.1-1211.
- Forster, "Intermolecular Energy Transfer and Fluorescence, *Annalen der Physik*" (1948) pp. 55-75.
- Gamoh, et al., 222 *Analytica Chimica Acta* 201 (1989).
- Gamsey, S. et al. 2007 "Boronic acid based bipyridinium salts as tunable receptors for monosaccharides and alpha-hydroxycarboxylates" *J Am Chem Soc* 129:1278-1286.
- Gamsey, Soya et al.: "Continuous glucose detection using boronic acid-substituted viologens in, fluorescent hydrogels: linker effects and extension to fiber optics" *Langmuir*, ACS, Washington, DC vol. 22, No. 21, Oct. 10, 2006, pp. 9067-9074 (XP002442273 ISSN: 0743-7463, compound (1) schemata 1,2 figure 1).
- Gehrich, J. L., D. W. Lubbers, et al. (1986). "Optical fluorescence and its application to an intravascular blood gas monitoring system." *IEEE TBio-med Eng BME-33*: 117-132.
- Glazer, "The Time-Dependent Specific Interaction of 4-(4'-Aminophenylazo)Phenylarsonic Acid with Subtilisins," 59 *Biochemistry* 996 (1968).
- Glazer, *Chemical Abstracts*, vol. 68, No. 23, Jun. 3, 1968, p. 111809 (total 3 pages).
- Guilbault, George E., "Practical Fluorescence" (1973), pp. 599-600.
- Hakkinen, Lajunen & Purokoski, A Potentiometric Study on the Complex Formation of Lactitol and Maltitol with Some Inorganic Oxyacids in Aqueous Solution, *Chemical Abstracts*, vol. 110, No. 83116f (1989).
- Hakkinen, Purokoski & Lajunen, A Potentiometric Study on the Complex Formation of Germanic Acid and Germanate Ion with Sugar Acids and Disaccharides in Aqueous Solution, *Chemical Abstracts*, vol. 105, No. 233265s (1986).

(56)

## References Cited

## OTHER PUBLICATIONS

- Hayashi, et al., "Fluorometric measurement of glycosylated albumin in human serum," 149 *Clinica Chimica Acta* 149 (1985), 13-19 Elsevier.
- Hirata O. et al. 2002 "Allosteric saccharide sensing by a phenylboronic-acids-appended 5,15-bis(triarylethynyl)porphyrin" *J Supramolecular Chemistry* 2:133-142.
- Hirshfeld, "Reabsorption Sensing in Fluorescence Spectroscopy," UCRL Abstract No. 89736 Abst, published by Pittsburg Conference on Scientific Instrumentation, Mar. 1984.
- Japanese First Office Action, re JP Application No. 2009-549167, dated Nov. 29, 2011.
- Japanese Office Action, re JP Application No. 2009-521962, dated Jul. 31, 2012.
- Japanese Office Action, re JP Application No. 2010-507712, dated Jun. 15, 2012.
- Kostov, Y., P. Harms, et al. (2001). "Low-cost microbio-reactor for high-throughput bioprocessing." *Biotechnol Bioeng* 72: 346-352.
- Lakowitz et al., "Optical sensing of glucose using phase-modulation fluorimetry," *Analytica Chimica Acta*, 271, (1993), 155-164.
- Lindner, et al., 473 *J. Chromatography* 227-240 (1989).
- Lutty, G. A. (1978). "The acute intravenous toxicity of stains, dyes, and other fluorescent substances." *Toxicol Pharmacol.* 44: 225-229.
- Meadows and Schultz, "Design, manufacture and characterization of an optical fiber glucose affinity sensor based on an homogeneous fluorescence energy transfer assay system," (1993) *Anal. Chim. Acta* 280: pp. 21-30.
- Medtronic, Features of the Guardian REAL-Time Continuous Glucose Monitoring System, Features that fit your diabetes management lifestyle, located at <http://www.minimed.com/products/guardian/features.html> on Aug. 28, 2007.
- Mosbach, *Methods in Enzymology*, vol. XLIV, 53 (1976).
- Niu C.G. et al. "Fluorescence ratiometric pH sensor prepared from covalently immobilized porphyrin and benzothioxanthene." 2005 *Anal. Bioanal. Chem.* 383(2):349-357.
- Offenbacher, H., O. S. Wolfbeis, et al. (1986). "Fluorescence optical sensors for continuous determination of near-neutral pH values." *Sensor Actuator* 9: 73-84.
- PCT International Search Report and Written Opinion re PCT/US2008/052204, dated May 27, 2008.
- PCT International Search Report and Written Opinion re PCT/US2008/053226, dated Oct. 15, 2008.
- PCT Partial Search Report re PCT/US2008/053226 dated Jun. 27, 2008.
- PCT International Search Report re PCT/US2007/074255 dated Jul. 8, 2008 in 3 pages.
- PCT Report on Patentability and Written Opinion re PCT/US2007/074255 dated Jan. 27, 2009 in 9 pages.
- PCT International Preliminary Report on Patentability and Written Opinion re PCT/US2008/053097 dated Aug. 11, 2009.
- PCT International Search Report re PCT/US2008/053097 dated Jun. 27, 2008.
- PCT International Preliminary Report and Written Opinion re PCT/US2009/040379 dated Oct. 19, 2010.
- PCT International Search Report (Declaration of Non-Establishment of ISR) and Written Opinion re PCT/US2009/040379 dated Aug. 4, 2009.
- PCT Report on Patentability and Written Opinion re PCT/US2008/063332 dated Nov. 19, 2009.
- PCT Partial International Search Report re PCT/US2008/063332 dated Oct. 20, 2008.
- PCT International Preliminary Report on Patentability re PCT/US2008/063330 dated Nov. 19, 2009.
- PCT International Search Report and Written Opinion re PCT/US2008/063330 dated Sep. 3, 2008.
- PCT International Preliminary Report on Patentability and Written Opinion re PCT/US2008/072359, dated Feb. 9, 2010.
- PCT International Search Report and Written Opinion re PCT/US2008/072359 dated Dec. 15, 2008.
- PCT Partial International Search Report re PCT/US2008/072359 dated Oct. 15, 2008.
- PCT International Search Report and Written Opinion re PCT/US2008/062303 dated Aug. 14, 2008.
- PCT International Search Report and Written Opinion re PCT/US2008/069855 dated Apr. 16, 2009.
- PCT Preliminary Report re PCT/US2008/084239 dated May 25, 2010.
- PCT International Search Report and Written Opinion re PCT/US2008/084239 dated Jan. 29, 2009.
- PCT International Search Report and Written Opinion re PCT/US2010/044761, dated Oct. 6, 2010.
- PCT International Search Report and Written Opinion re App. No. PCT/US10/61169, dated Mar. 1, 2011.
- PCT International Preliminary Report on Patentability and Written Opinion re PCT Application No. PCT/US20101061169, dated Jun. 19, 2012.
- PCT International Search Report and Written Opinion re PCT/US2010/037502, dated Aug. 6, 2010.
- PCT International Preliminary Report on Patentability re PCT/US20101037502, dated Dec. 6, 2011.
- PCT International Search Report and Written Opinion re PCT App. No. PCT/US 10/50910, dated Dec. 3, 2010.
- PCT International Preliminary Report on Patentability and Written Opinion re PCT/US 10/50910 date of Issuance Apr. 3, 2012/date of mailing Apr. 12, 2012.
- PCT International Search Report and Written Opinion re App. No. PCT/US 10/61163, dated Mar. 9, 2011.
- PCT International Search Report and Report on Patentability re PCT Application No. PCT/US20101061163, dated (mailed) Jun. 28, 2012.
- PCT International Search Report and Written Opinion re App. No. PCT/US10/61173, dated Feb. 28, 2011.
- PCT International Preliminary Report on Patentability and Written Opinion re PCT Application No. PCT/US20101061173, dated Jun. 19, 2012.
- PCT International Search Report and Written Opinion in App. No. PCT/US2011/028222, dated May 6, 2011, in 30 pages.
- PCT Search Report and Written Opinion re PCT/US2011/023939, dated Jul. 27, 2011.
- PCT International Search Report and Written Opinion re PCT/US2011/034167, mailed Jul. 29, 2011.
- Purokoski, Lajunen & Hakkinen, A Potentiometric Study on the Complex Formation of Arsenious Acid, Arsenite Ion, Telluric Acid and Tellurate Ion with Sugar Acids and Disaccharides in Aqueous Solution, *Chemical Abstracts*, vol. 107, No. 122178n (1987).
- Roy, et al., *J. Inorg. Nucl. Chem.*, 106 (1957).
- Schulman, S. G., S. Chen, et al. (1995). "Dependence of the fluorescence of immobilized 1-hydroxypyrene-3,6,8-trisulfonate on solution pH: extension of the range of applicability of a pH fluorosensor." *Anal Chim Acta* 304: 165-170.
- Seitz, "Chemical Sensors Based on Fiber Optics," *Analytical Chemistry*, vol. 56, pp. 16a-34a, 1984.
- Sharrett, Z. et al. 2008 "Boronic acid-appended bis-viologens as a new family of viologen quenchers for glucose sensing" *Tetrahedron Letters* 49:300-304.
- Snyder, et al., "The Preparation of Some Azo Boronic Acids," 70 *J. Am. Chem. Soc.* 232 (1948).
- Song, A., S. Parus, et al. (1997) "High-performance fiber-optic pH microsensors for practical physiological measurements using a dual-emission sensitive dye." *Analytical Chemistry* 69: 863-867.
- Streitwieser, Jr. & Heathcock, *Introduction to Organic Chemistry* (1976).
- Sturdevant, M. F.: "How Sterilization Changes Long-Term Resin Properties", *Plastics Engineering*, Jan. 1991, pp. 27-32.
- Suri, J. T. et al. 2003 "Continuous glucose sensing with a fluorescent thin-film hydrogel" *Angew Chem Int Ed* 42:5857-5859.
- The Immunoassay Handbook, pp. 1-618, ed. David Wild, Macmillan Press, 1994, United Kingdom.
- Turner N.G. et al. "Determination of the pH Gradient Across the Stratum Corneum." 1998 *J. Investig. Dermatol. Symp. Proc.* Aug. 3(2):110-3.
- Udenfreund, "Fluorescence Assay in Biology and Medicine" (1962) pp. 108-109.

(56)

**References Cited**

OTHER PUBLICATIONS

Van Kempfen & Kreuzer, "A Single-Unit Carbon Dioxide Sensing Microelectrode System," *Respiration Physiology*, (1975), 23, 371-379.

Vermeer, et al., 37 *Tetr. Letters* 3255 (1970).

Wang, D. et al. 2001 "Photoluminescence quenching of conjugated macromolecules by bipyridinium derivatives in aqueous media: charge dependence" *Langmuir* 17:1262-1266.

Wolfbeis, O. S., E. Fuerlinger, et al. (1983). "Fluorimetric analysis. I. Study on fluorescent indicators for measuring near neutral ('physiological') pH values." *Fresenius' Z. Anal. Chem.* 314 (2): 119-124.

Xu, Z., A. Rollins, et al. (1998) "A novel fiber-optic pH sensor incorporating carboxy SNAFL-2 and fluorescent wavelength-ratiometric detection" *Journal of Biomedical Materials Research* 39:9-15.

Zhang, S., S. Tanaka, et al. (1995). "Fibre-optical sensor based on fluorescent indicator for monitoring physiological pH values." *Med Biol Eng Comput* 33: 152-156.

Zhujun, Z. and W. R. Seitz (1984). "A fluorescence sensor for quantifying pH in the range from 6.5 to 8.5." *Analytical Chimica Acta* 160:47-55.

Zhujun, Z., et al. (1984). *Analytical Chimica Acta* 160:305-309.

\* cited by examiner

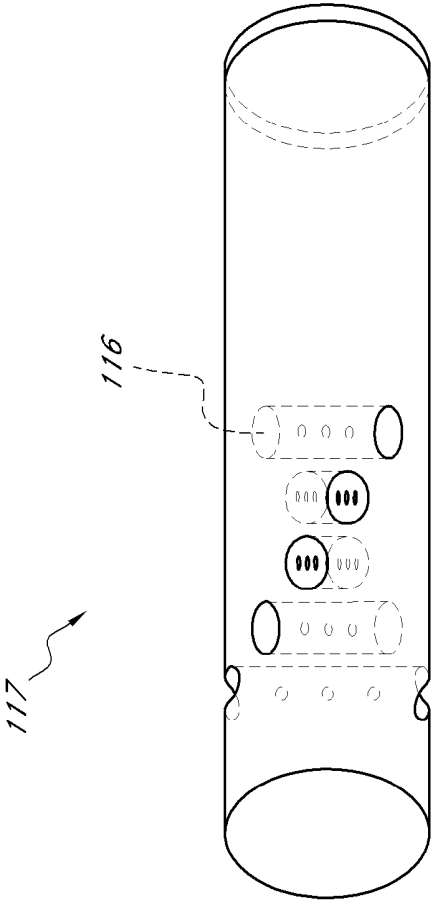


FIG. 1A

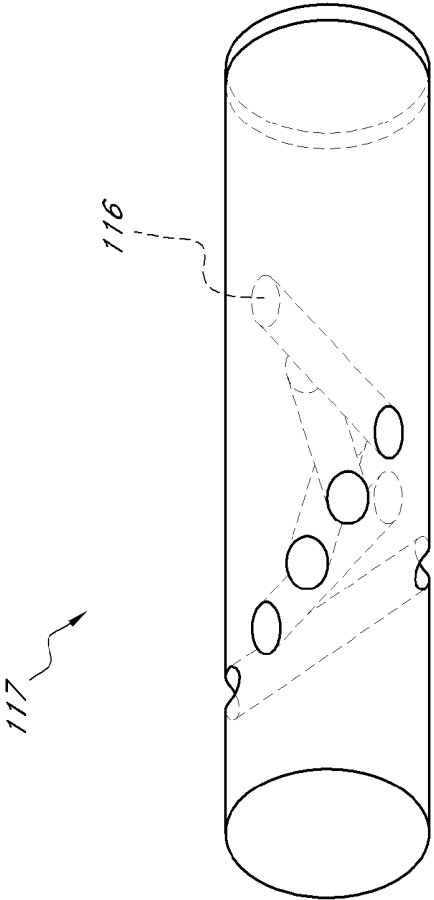


FIG. 1B

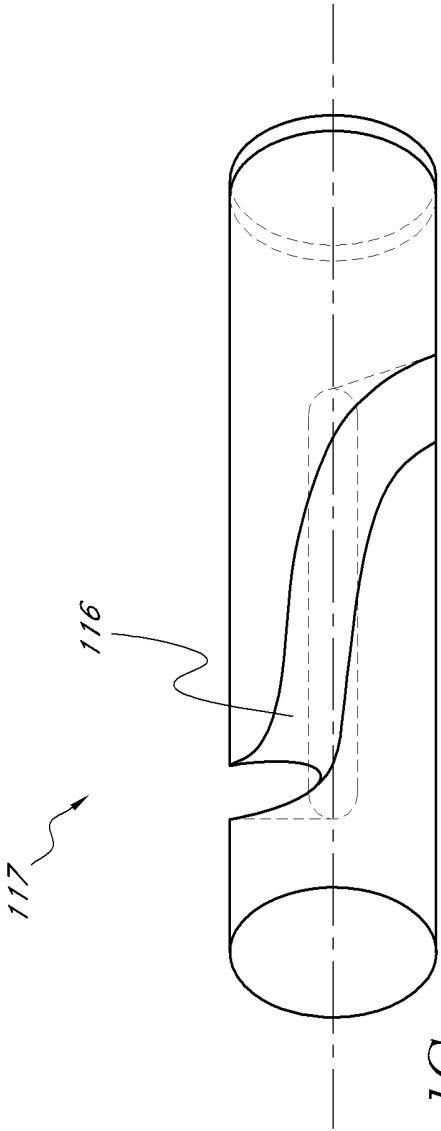


FIG. 1C

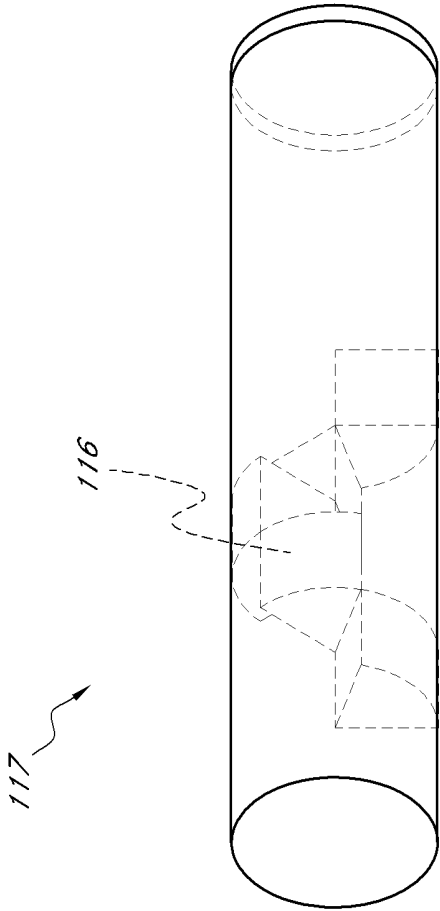


FIG. 1D

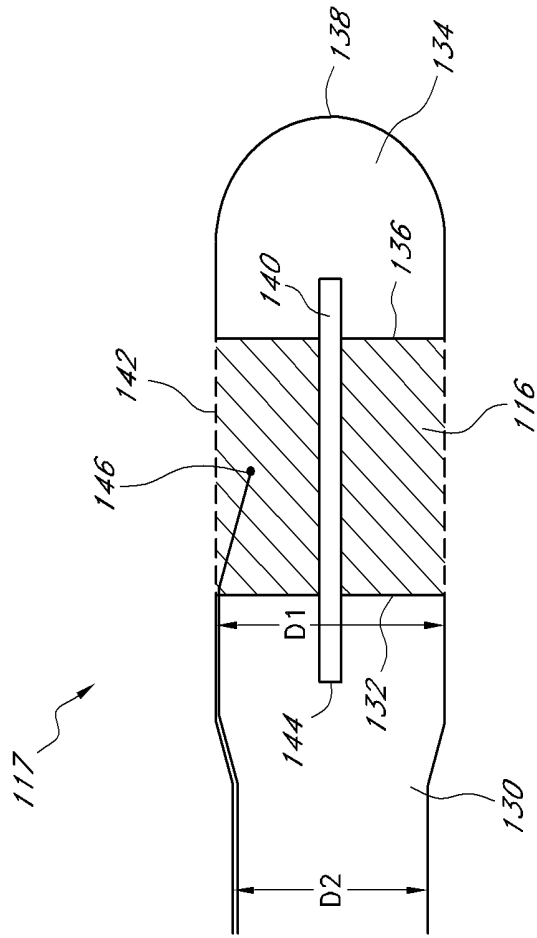
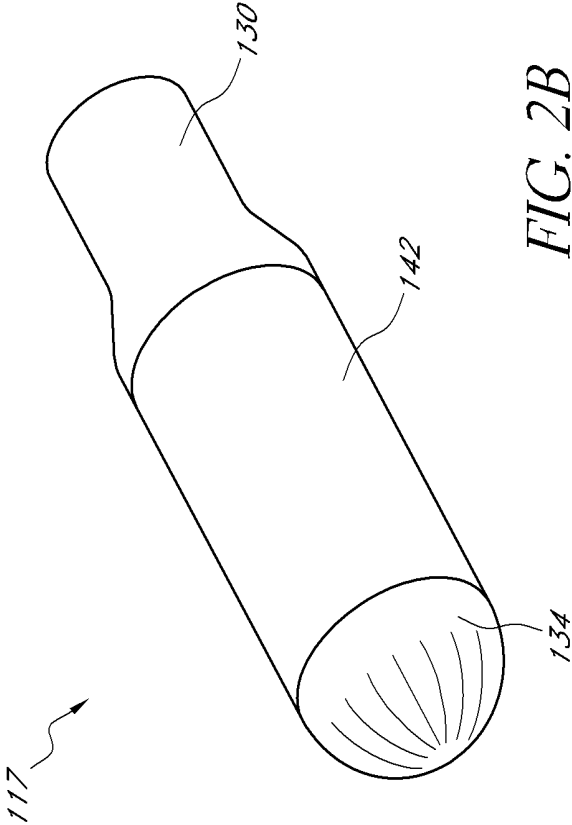


FIG. 2A



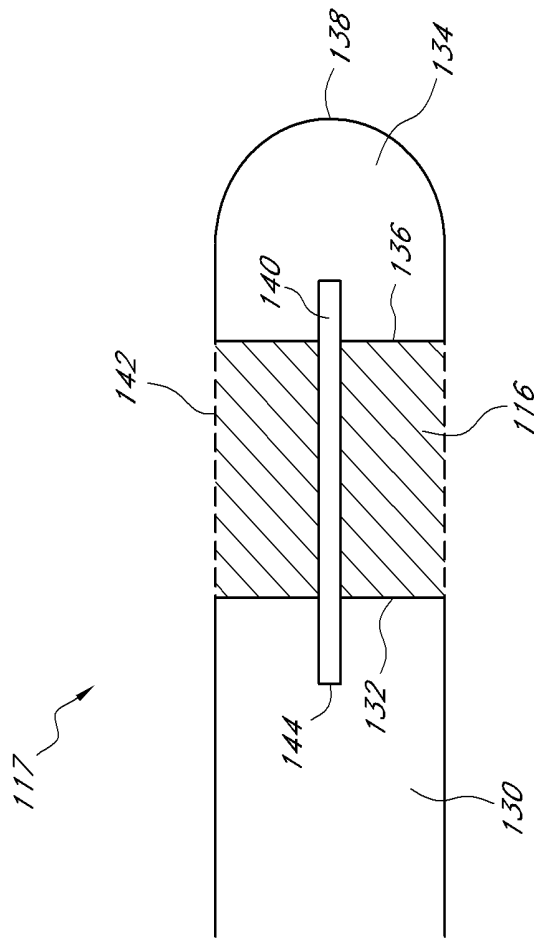


FIG. 3A

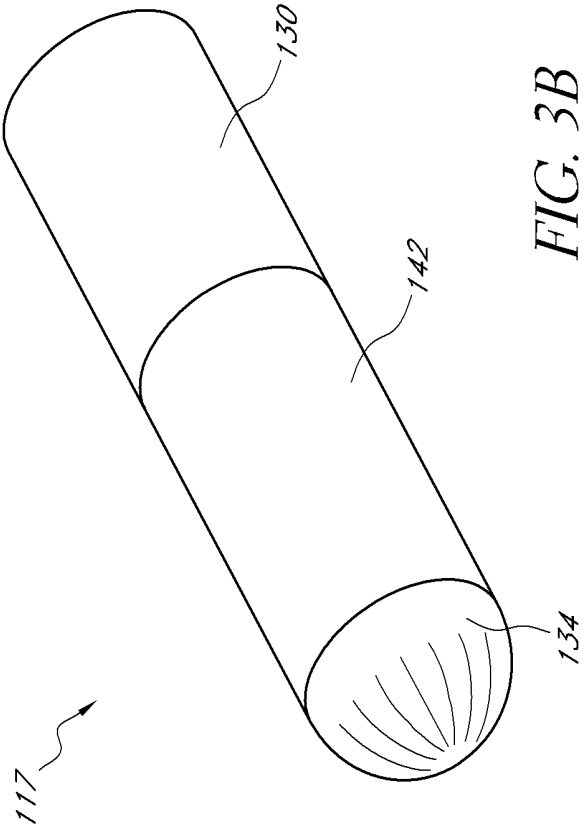


FIG. 3B

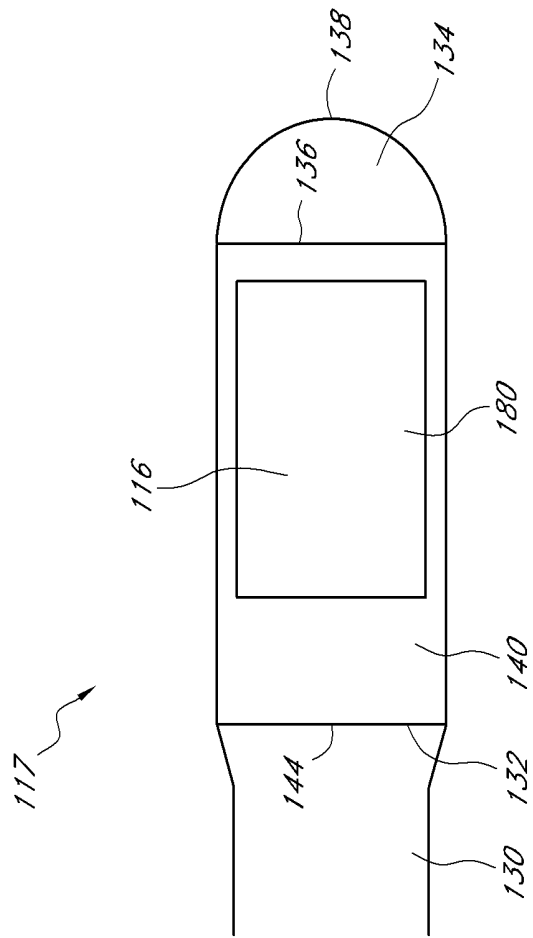


FIG. 4

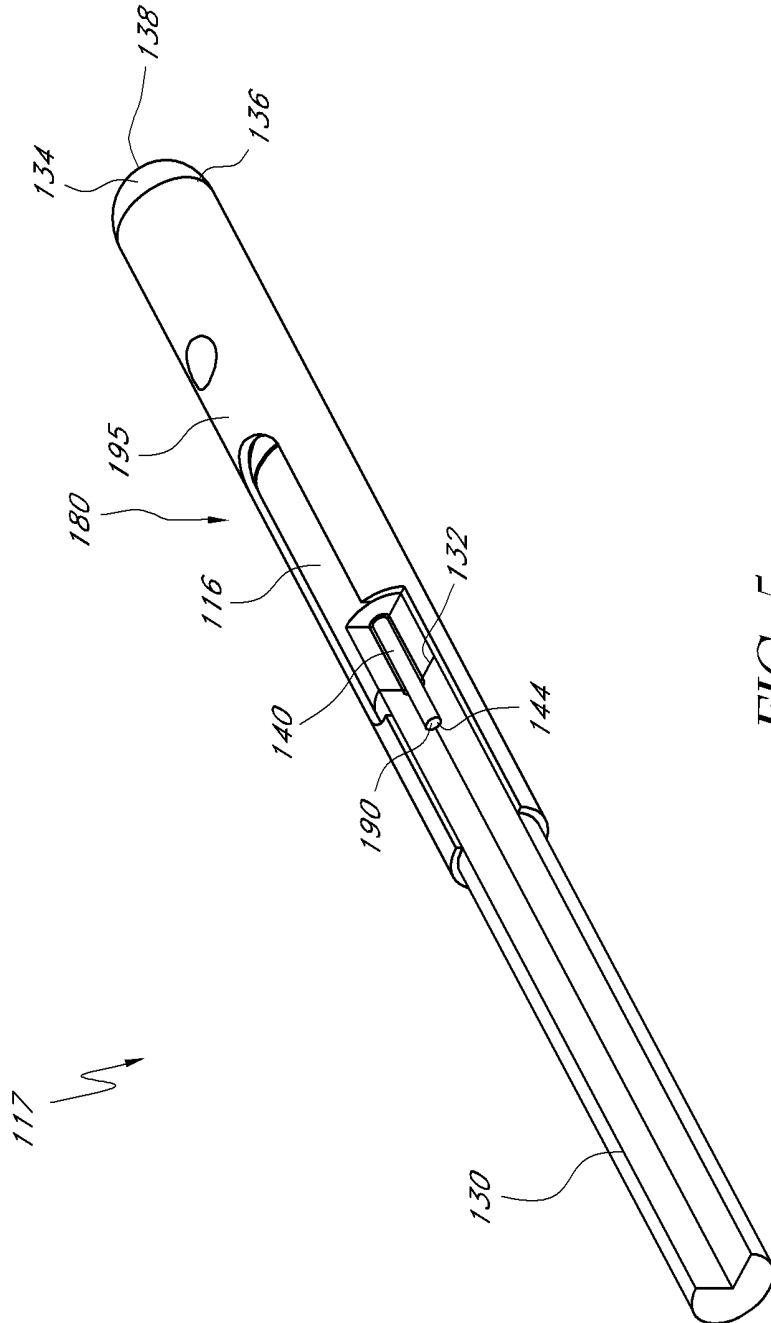


FIG. 5

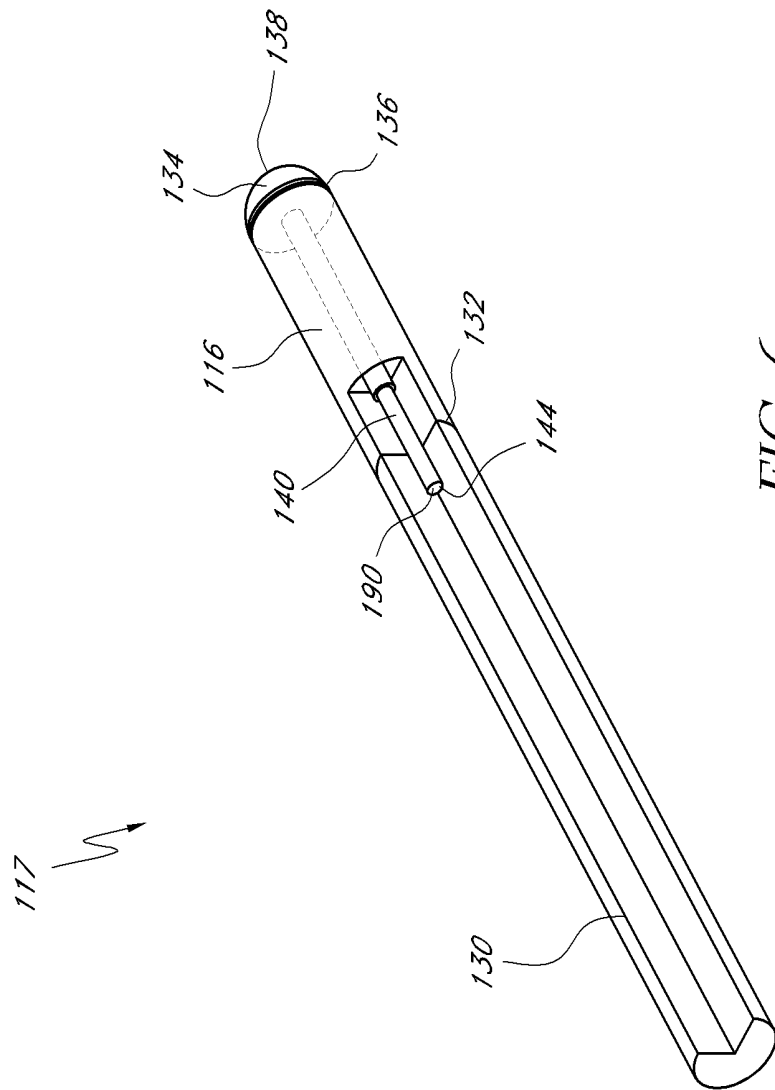


FIG. 6

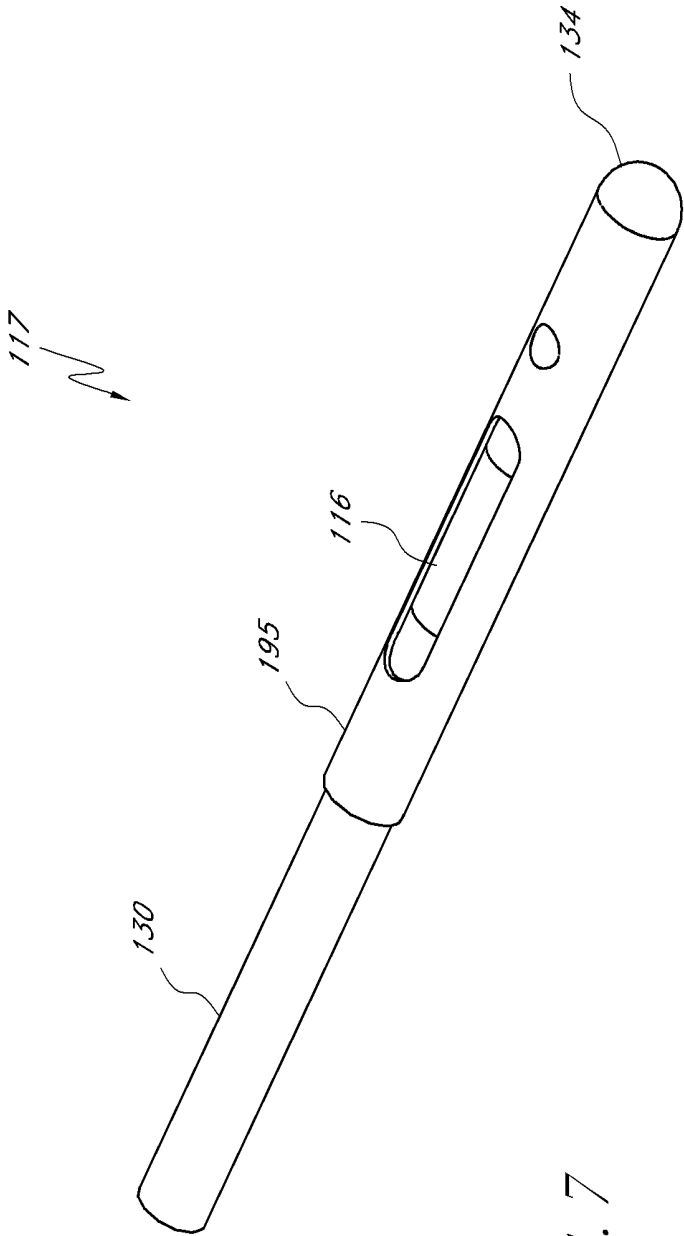


FIG. 7

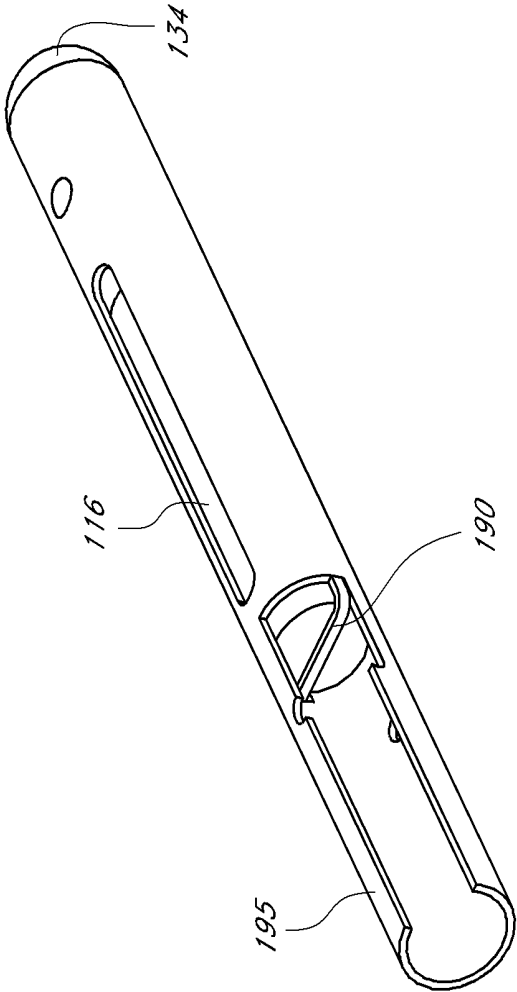
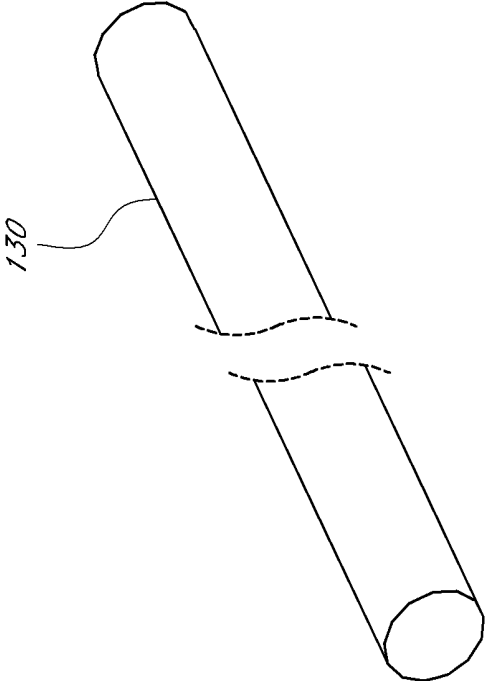
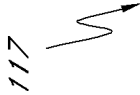


FIG. 8



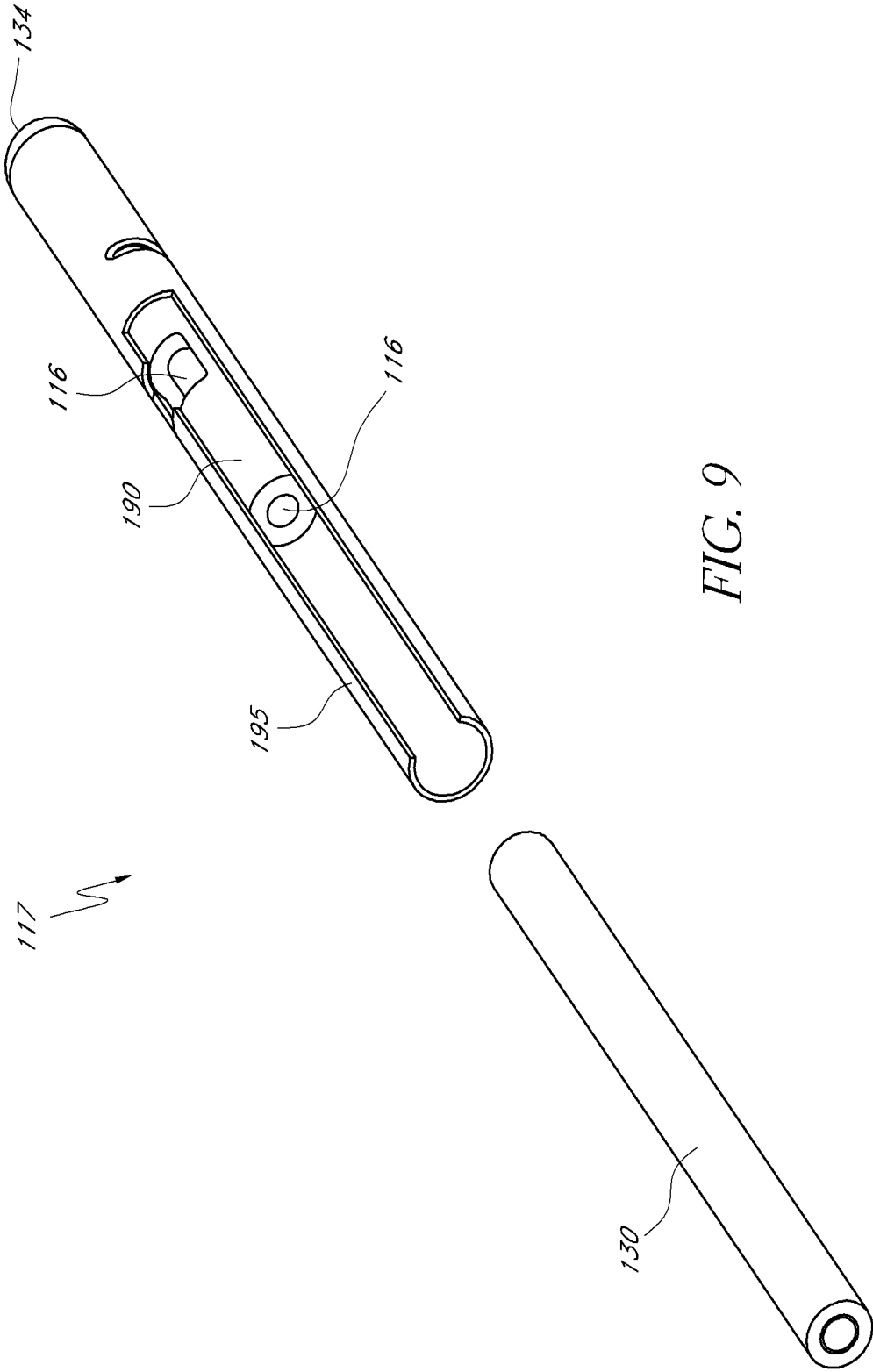


FIG. 9

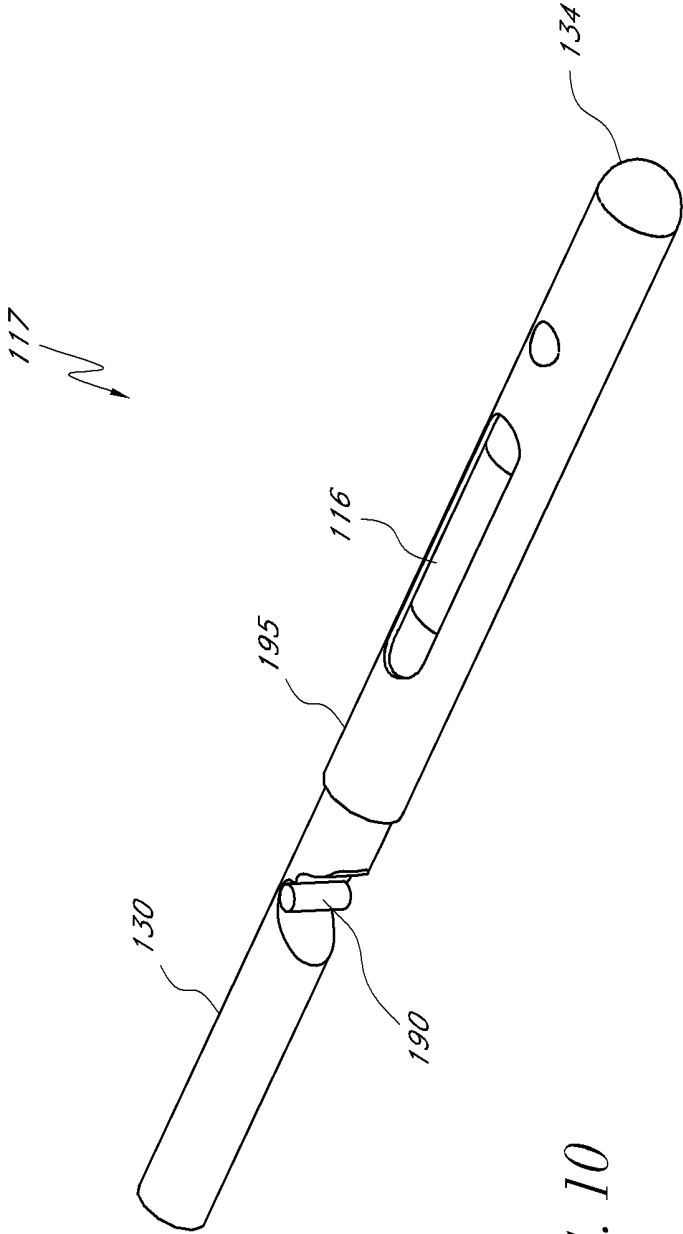


FIG. 10

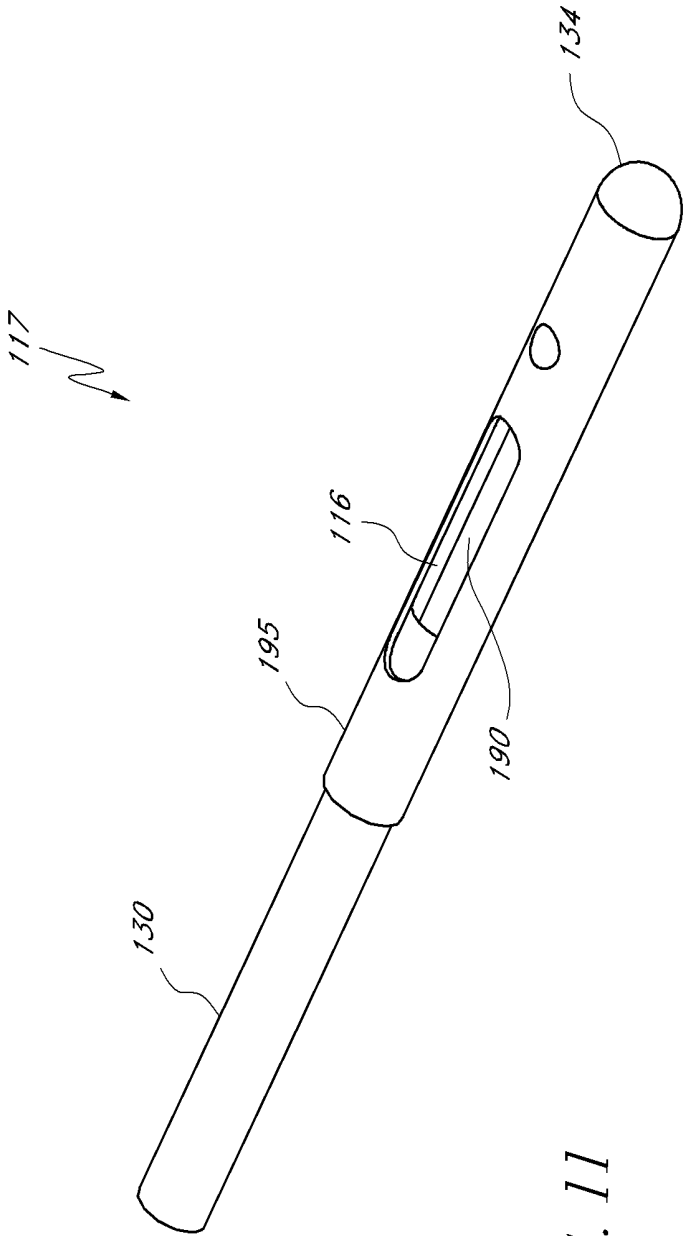


FIG. 11

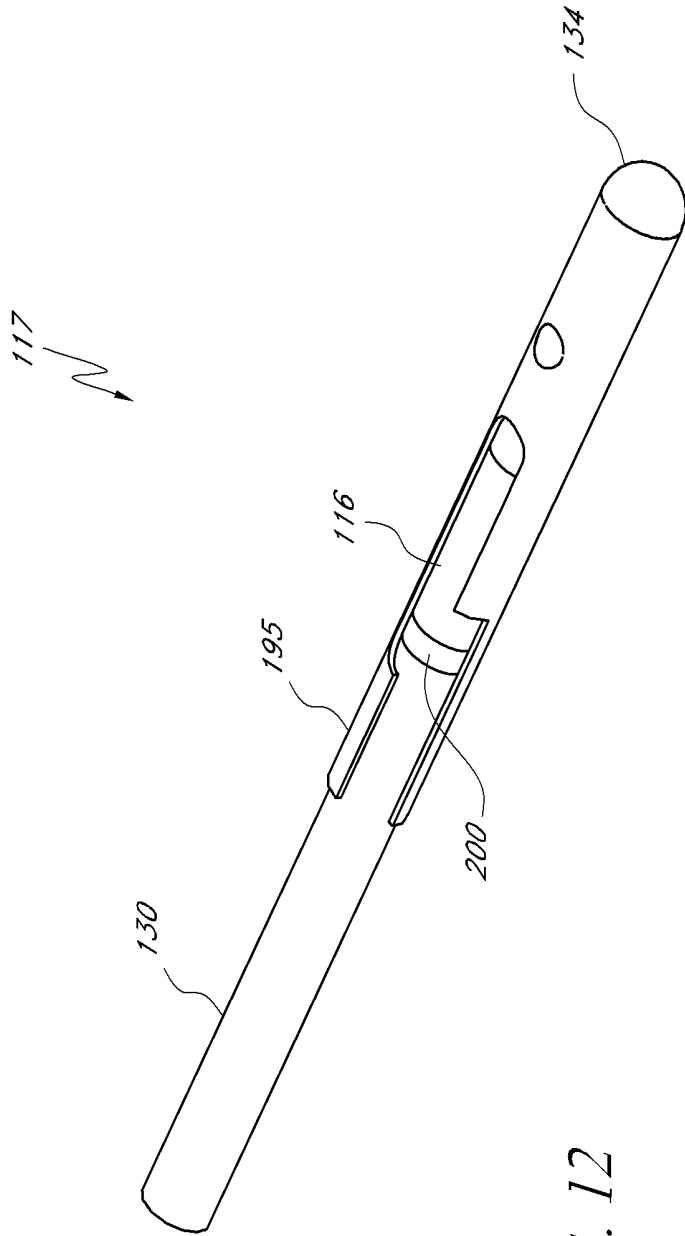


FIG. 12

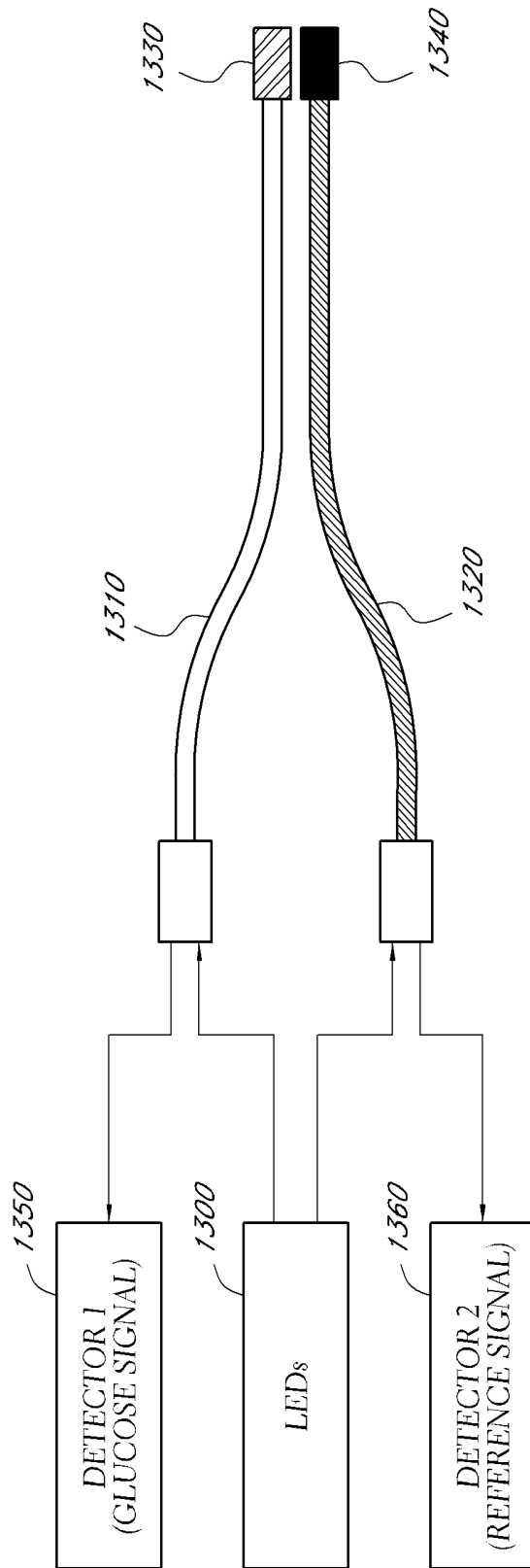


FIG. 13

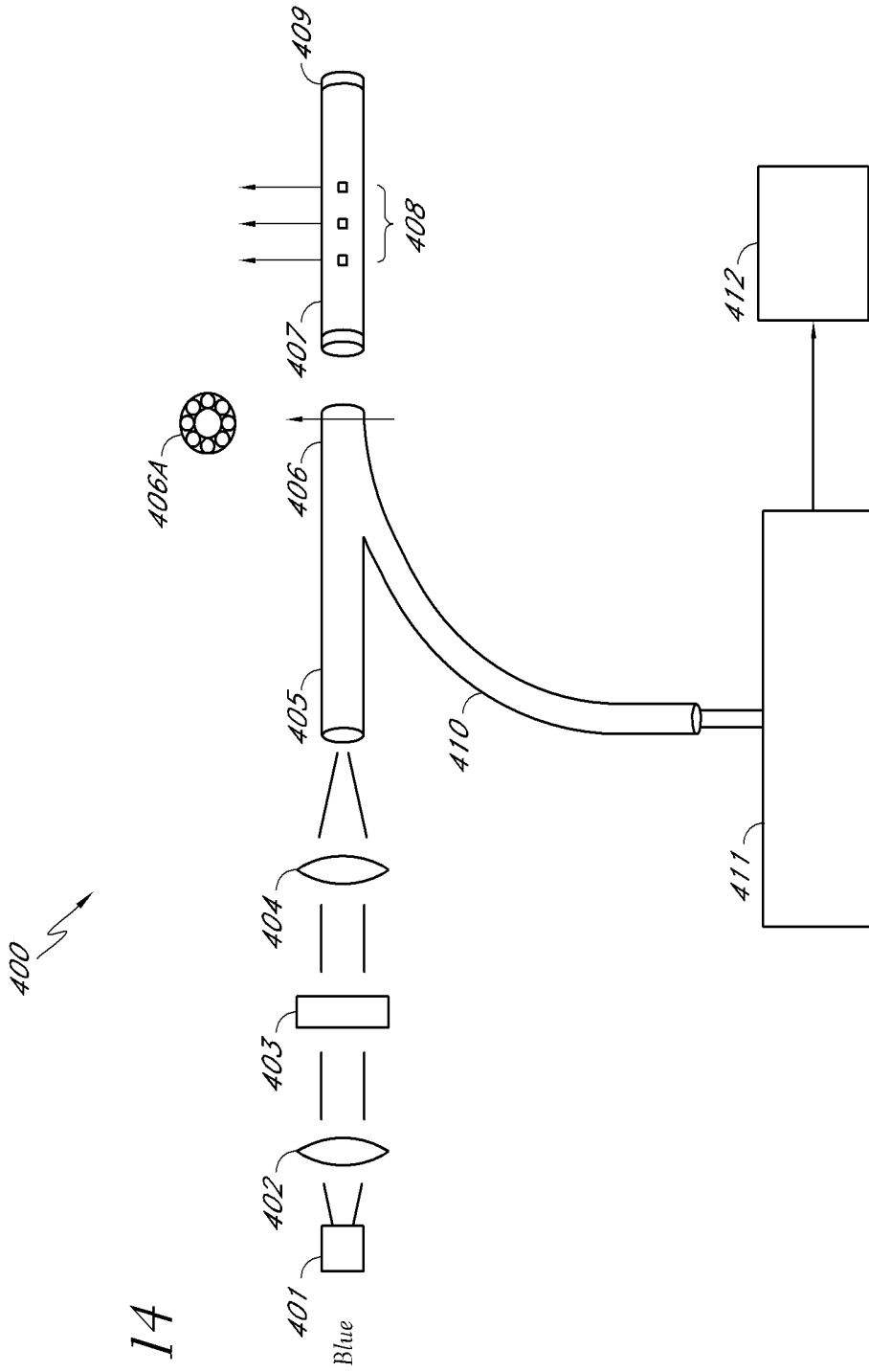


FIG. 14

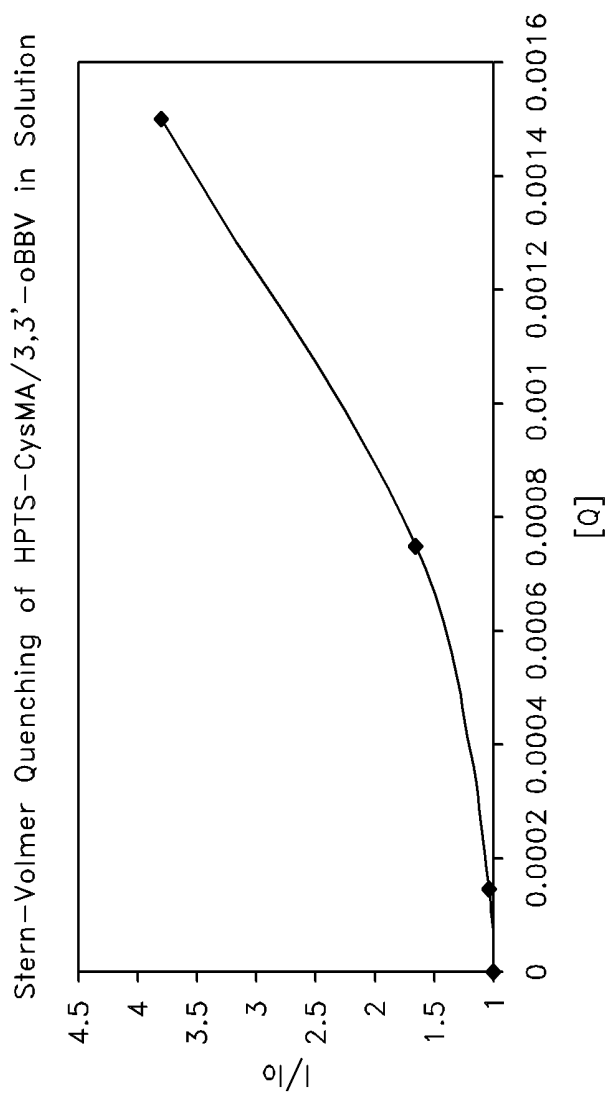


FIG. 15

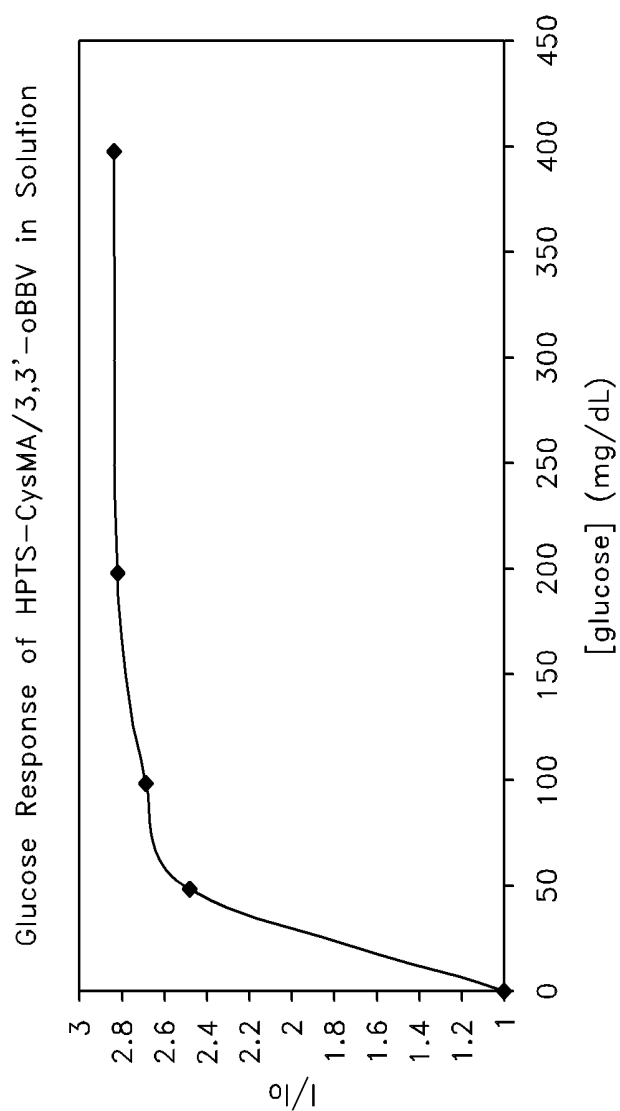


FIG. 16

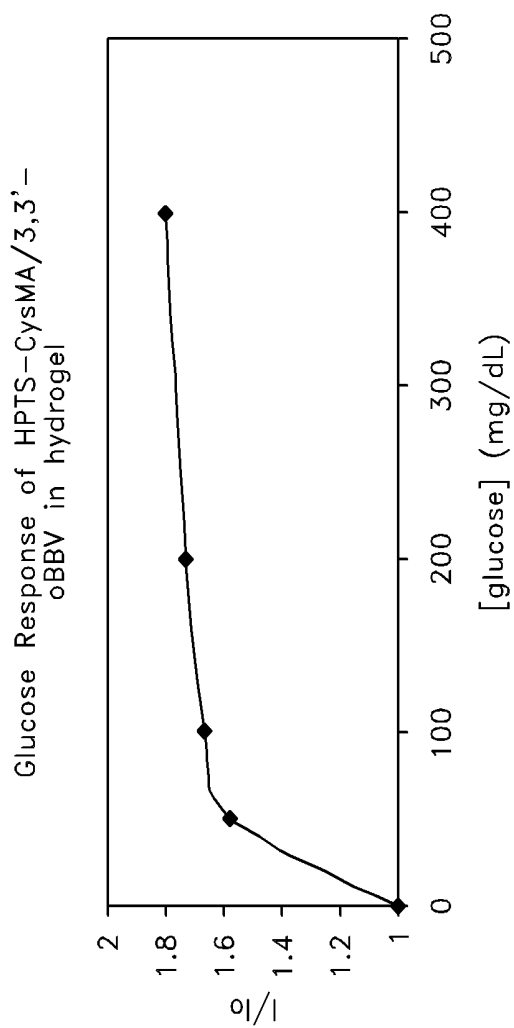


FIG. 17

## OPTICAL SENSOR CONFIGURATION FOR RATIOMETRIC CORRECTION OF GLUCOSE MEASUREMENT

### INCORPORATION BY REFERENCE TO ANY PRIORITY APPLICATIONS

Any and all applications for which a foreign or domestic priority claim is identified in the Application Data Sheet as filed with the present application are hereby incorporated by reference under 37 CFR 1.57.

### BACKGROUND OF THE INVENTION

#### 1. Field of the Invention

Embodiments of the invention are directed to an optical sensor for detecting an analyte, preferably glucose. In preferred embodiments, the sensor comprises an optical fiber having a fluorescence chemistry disposed along the distal region of the fiber, more preferably located between the distal end of the fiber and an atraumatic tip.

#### 2. Description of the Related Art

Hyperglycemia and insulin resistance are common in critically ill patients, even if such patients have not previously had diabetes. In these situations, glucose levels rise in critically ill patients thereby increasing the risk of damage to a patient's organs. Further, studies have shown that normalization of blood glucose levels with insulin therapy improves the prognosis for such patients, thereby decreasing mortality rates.

More recent scientific evidence confirms that dramatic improvements in the clinical outcome of hospitalized Intensive Care Unit (ICU) patients can result from tight therapeutic control of blood glucose to normal ranges. These studies indicate that Tight Glycemic Control (TGC) of ICU patients may reduce mortality by as much as 40%, and significantly lower complication rates. In these situations, it is necessary to accurately, conveniently and continuously monitor blood sugar in a real-time device specifically designed to meet the challenging needs of the ICU environment. Researchers at Johns Hopkins University estimate that TGC can save as many as 150,000 lives and reduce U.S. healthcare costs by as much as \$18 billion annually.

Performing TGC requires continuous and accurate monitoring of a patient's blood glucose levels. Thus, there is a need for a real-time glucose monitoring system that is adapted to meet the needs of ICU patients.

### SUMMARY OF THE INVENTION

A sensor for detecting an analyte concentration in a blood vessel is disclosed in accordance with an embodiment of the invention. The sensor comprises: an optical fiber with proximal and distal ends; an atraumatic tip portion with proximal and distal ends, wherein the proximal end of the atraumatic tip portion is separated from the distal end of the optical fiber, such that a gap exists between the atraumatic tip portion and the optical fiber; a rod with proximal and distal ends, wherein the proximal end of the rod is attached to the distal end of the optical fiber, and wherein the distal end of the rod is attached to the proximal end of the atraumatic tip portion, such that the rod traverses the gap and couples the optical fiber to the atraumatic tip portion; a chemical indicator system capable of generating an emission light signal in response to an excitation light signal, wherein the intensity of the emission light signal is related to the analyte concentration, and wherein the chemical indicator system is disposed within the gap; and a

selectively permeable membrane disposed over the gap, wherein the sensor is sized for deployment within the blood vessel.

In one variation to the analyte sensor, the chemical indicator system is immobilized within the gap by a hydrogel. In another variation, the sensor further comprises a temperature sensor. The optical fiber preferably has a diameter of between about 0.005 inches and about 0.020 inches. In another variation, the sensor further comprises a reflective region. Preferably, the reflective region comprises a reflective surface of the proximal end of the rod. In one embodiment, the rod may be attached to the optical fiber and atraumatic tip portion by heating. In another embodiment, the rod may be attached to the optical fiber by a reflective or optically clear adhesive.

In variations to the sensor, the shape of the distal end of the atraumatic tip portion may be configured to reduce trauma within the blood vessel. In various embodiments, the shape of the distal end of the atraumatic tip portion may be selected from the group consisting of hemispherical, parabolic, and elliptical. In another variation, the distal end of the atraumatic tip portion is flexible. In another variation, the distal end of the atraumatic tip portion is deformable. The distal end of the atraumatic tip portion may be formed from at least one material selected from the group consisting of plastics, polymers, gels, metals and composites.

The rod may be formed from at least one material selected from the group consisting of metal, metal alloy, plastic, polymer, ceramic, and composite material. In a preferred variation, the rod is formed from stainless steel, titanium, or Nitinol. In one embodiment, the rod is cylindrical. Preferably, the rod diameter is between about 0.002 inches and about 0.010 inches. The rod may be flexible in some embodiments. The rod is stiffer than the optical fiber in some embodiments. In such embodiments, the rod is preferably sufficiently stiff to prevent flexing of the sensor along the gap.

A sensor for detecting an analyte concentration in a blood vessel is disclosed in accordance with another embodiment of the present invention. The sensor comprises: an optical fiber with proximal and distal ends; an atraumatic tip portion with proximal and distal ends, wherein the proximal end of the atraumatic tip portion is separated from the distal end of the optical fiber, such that a gap exists between the atraumatic tip portion and the optical fiber; a hypotube with proximal and distal ends, wherein the proximal end of the hypotube is attached to the distal end of the optical fiber, and wherein the distal end of the hypotube is attached to the proximal end of the atraumatic tip portion, such that the hypotube traverses the gap and couples the optical fiber to the atraumatic tip portion, wherein the hypotube comprises at least one window that opens onto the gap; a chemical indicator system capable of generating an emission light signal in response to an excitation light signal, wherein the intensity of the emission light signal is related to the analyte concentration, and wherein the chemical indicator system is disposed within the gap; and a selectively permeable membrane disposed over the at least one window, wherein the sensor is sized for deployment within the blood vessel. In preferred embodiments, the chemical indicator system is immobilized by a hydrogel within the cavity formed within the hypotube. In further preferred embodiments of the sensor with hypotube, a reflective member is disposed within the sensor. In further preferred embodiments of the sensor with hypotube, a fluorescent member is disposed within the sensor.

A sensor for detecting an analyte concentration in a blood vessel is disclosed according to another embodiment of the present invention. The sensor comprises: an optical fiber with proximal and distal ends; an atraumatic tip portion with

proximal and distal ends, wherein the proximal end of the atraumatic tip portion is separated from the distal end of the optical fiber, such that a gap exists between the atraumatic tip portion and the optical fiber; a cage connecting the optical fiber and atraumatic tip portion, wherein the optical fiber is at least partially enclosed within the cage, and wherein the cage has at least one window; a chemical indicator system disposed within the cage, wherein the chemical indicator system is adjacent the window and is separated from analyte by a selectively permeable membrane, and wherein the chemical indicator system is capable of generating an emission light signal in response to an excitation light signal, wherein the intensity of the emission light signal is related to the analyte concentration; and a reference material, wherein the reference material is configured to either reflect a portion of the excitation light signal before the excitation light signal enters the chemical indicator system or to return a second emission light signal, wherein the intensity of the second emission light signal is not related to the analyte concentration.

In another embodiment, a method for measuring glucose concentration is provided. The method comprises transmitting a first light in a first direction through an optical fiber to a glucose sensor, where the glucose sensor comprises a hydrogel cavity having a fluorophore system. At least a portion of the first light is reflected off a reflective surface of the glucose sensor as a second light in a second direction opposite the first direction. A third light is emitted in the second direction. The third light results from the chemical indicator system fluorescing. The method further comprises calculating the glucose concentration, where the glucose concentration is determined by the ratio of the emitted third light to the reflected second light. The ratio is independent of the intensity of the first light.

In another embodiment, a method for measuring glucose concentration is provided. The method comprises transmitting a first light in a first direction through an optical fiber to a glucose sensor, where the glucose sensor comprises a hydrogel cavity having a fluorophore system which is sensitive to glucose as well as a second fluorophore which is glucose insensitive. A second light from the glucose sensitive fluorophore is emitted in a second direction opposite the first direction. A third light from the glucose insensitive fluorophore is also emitted in the second direction. The method further comprises calculating the glucose concentration, where the glucose concentration is determined by the ratio of the emitted second light to the emitted third light. The ratio is independent of the intensity of the first light.

In another embodiment, a system for measuring glucose is provided. The system comprises at least one light source, at least one optical fiber coupled to the light source, any one of the glucose sensors described above coupled to the optical fiber, and a data processing device coupled to the glucose sensor.

In another embodiment, a method for manufacturing a glucose sensor is provided. The method comprises inserting a first end of a rod into an optical fiber, inserting a second end of the rod into an atraumatic tip, surrounding the rod with a hydrogel cavity, and enclosing the hydrogel cavity with a selectively permeable membrane.

In another embodiment, a method for manufacturing a glucose sensor is provided. The method comprises cutting a window in a hypotube, contacting an optical fiber with a first end of the hypotube, and heating the optical fiber, such that the optical fiber swells to fully contact the first end of the rod.

In another embodiment, a sensor for detecting an analyte concentration in a blood vessel is provided. The sensor comprises an optical fiber with a proximal end and a distal end.

The distal end of the optical fiber comprises a glucose sensing hydrogel. The glucose sensing hydrogel comprises a first fluorophore, a quencher, and at least one glucose receptor. A reference fiber is adjacent the optical fiber and has a proximal end and a distal end. The distal end of the reference fiber comprises a reference material. The reference material comprises a second fluorophore. A light emitting diode is operably coupled to the glucose fiber and the reference fiber. The light emitting diode sends an excitation light to the glucose fiber and the reference fiber. A glucose signal detector is operatively coupled to the glucose fiber. The glucose signal detector receives a first fluorescent light from the glucose fiber. A reference signal detector is operatively coupled to the reference fiber. The reference signal detector receives a second fluorescent light from the reference fiber.

#### BRIEF DESCRIPTION OF THE DRAWINGS

FIG. 1A shows a glucose sensor having a series of holes that form a helical configuration.

FIG. 1B shows a glucose sensor having a series of holes drilled or formed at an angle.

FIG. 1C shows a glucose sensor having at least one spiral groove.

FIG. 1D shows a glucose sensor having a series of triangular wedge cut-outs.

FIG. 2A shows a cross-sectional view of one embodiment of a glucose sensor having a cavity in the distal portion of the sensor and a temperature probe.

FIG. 2B shows a perspective view of the glucose sensor shown in FIG. 2.

FIG. 3A shows a cross-sectional view of another embodiment of a glucose sensor having a cavity in the distal portion of the sensor.

FIG. 3B shows a perspective view of the glucose sensor shown in FIG. 4.

FIG. 4 shows a cross-sectional view of another embodiment of a glucose sensor having a window opening to a cavity in the distal portion of the sensor.

FIG. 5 shows a cross-sectional view of another embodiment of a glucose sensor having a cavity in a distal portion of the sensor enclosed within a cage and an additional reference material.

FIG. 6 shows a cross-sectional view of another embodiment of a glucose sensor having a cavity in a distal portion of the sensor and an additional reference material.

FIG. 7 shows a cross-sectional view of another embodiment of a glucose sensor having a cavity in a distal portion of the sensor enclosed within a cage and a reference material extending to the atraumatic tip.

FIG. 8 shows a cross-sectional view of another embodiment of a glucose sensor having a cavity in a distal portion of the sensor enclosed within a cage and a reference material as a bar extending across the diameter of the cage.

FIG. 9 shows a cross-sectional view of another embodiment of a glucose sensor having a cavity in a distal portion of the sensor enclosed within a reference material, further enclosed within a cage.

FIG. 10 shows a cross-sectional view of another embodiment of a glucose sensor having a cavity in the distal portion of the sensor enclosed within a cage and a reference material as a bar embedded within the optical fiber.

FIG. 11 shows a cross-sectional view of another embodiment of a glucose sensor having a cavity and a reference material side-by-side in the distal portion of the sensor enclosed within a cage.

FIG. 12 shows a cross-sectional view of another embodiment of a glucose sensor having a cavity in the distal portion of the sensor enclosed within a cage and a translucent reference material between the optical fiber and cavity.

FIG. 13 shows a schematic view of another embodiment of a glucose sensor having a glucose sensing optical fiber adjacent to a reference optical fiber.

FIG. 14 shows a glucose measurement system comprising one excitation light source, a single exciter-dual emitter fluorophore system, and a microspectrometer and/or spectrometer.

FIG. 15 shows the Stern-Volmer quenching of HPTS-CysMA/3,3'-oBBV in solution.

FIG. 16 shows the glucose response of HPTS-CysMA/3,3'-oBBV in solution.

FIG. 17 shows the glucose response of HPTS-CysMA/3,3'-oBBV in hydrogel.

#### DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT

Various embodiments of optical systems and methods are disclosed herein for determining blood glucose concentrations. The various embodiments preferably share at least two features. First, they involve exciting a chemical indicator system with an excitation light signal and measuring the emission light signal of the indicator system, wherein the indicator system is in contact with the blood and comprises a fluorescent dye operably coupled to a glucose binding moiety—such that the emission light signal generated by the indicator system upon excitation is related to the blood glucose concentration. Second, they involve correcting the blood glucose concentration measurements from the indicator system for potential artifacts due to the optical system, which artifacts are unrelated to the blood glucose concentration. The correction is performed by ratiometric analysis. More particularly, the ratio of emission light signal to a second light signal that is propagated through the optical system, e.g., the excitation light signal or a separate reference light signal, is used for correcting any non-glucose related contributions of the optical system. Where the excitation light signal is used for the ratiometric correction, the sensor preferably includes a reflective surface, e.g., a mirror, located somewhere along the sensor, such that at least a portion of the excitation light which has passed through the optical system is reflected back to a detector. Where a separate reference light signal is used, the reference light signal may either be: (1) generated by a separate light source and reflected back to a detector, or (2) generated as a separate emission light signal from a separate dye disposed somewhere along the sensor. Thus, a glucose sensor in accordance with preferred embodiments of the present invention will comprise either a reflective surface or a second dye adapted to emit a reference light signal.

Various structural configurations have been proposed for holding a chemical indicator system in a position, which is: (1) exposed to the blood, (2) disposed within the excitation light path, and (3) for exposing a chemical indicator system to the blood and, for introducing to the indicator system an excitation light signal, for detecting an emission light signal from the indicator system, and for enabling ratiometric correction of glucose determinations for artifacts of the system optics; see in particular 2008/0188725. More particularly, aspects of the present invention relate to improvements and alternative embodiments for generating a reference light signal (as discussed in 2008/0188725), either through various mirror/reflective surface configurations adapted to return a portion of the excitation light signal back to a detector or

through generating a separate emission light signal from a separate dye. Aspects of the present invention relate to new and improved configurations for disposing a chemical indicator system within an interrogation light path, wherein the sensor is more robust and exhibits improved patient tolerance.

Optical glucose sensors, such as those described in U.S. Patent Publ. Nos. 2008/0188722, 2008/0188725, 2008/0187655, 2008/0305009, 2009/0018426, 2009/0018418, and co-pending U.S. patent application Ser. Nos. 11/296,898, 12/187,248, 12/172,059, 12/274,617 and 12/424,902 (each of which is incorporated herein in its entirety by reference thereto) typically employ a chemical indicator system disposed at the distal end of an optical fiber, wherein the indicator system is maintained in contact with the blood, such that an excitation light signal sent distally down the fiber causes the chemical indicator system to emit a light signal related to the concentration of glucose.

In certain embodiments, an optical glucose measurement system is disclosed for measuring glucose concentration in blood using one or more glucose-sensing chemical indicator systems. Such indicator systems preferably comprise a fluorophore operably coupled to a glucose binding moiety. Preferably, the glucose binding moiety acts as a quencher with respect to the fluorophore (e.g., suppresses the fluorescent emission signal of the fluorophore in response to excitation light when it associates with the fluorophore). In preferred embodiments, as the glucose binding moiety binds glucose (e.g., as glucose concentrations rise), it dissociates from the fluorophore, which then generates a fluorescent emission signal upon excitation. Accordingly, in such embodiments, the higher the glucose concentration, the more glucose bound by the binding moiety, the less quenching, and the higher the fluorescence intensity of the fluorophore upon excitation.

In certain embodiments, the optical glucose measurement system measures glucose concentrations intravascularly and in real-time through the use of such chemical indicator systems. In certain embodiments, the glucose-sensing chemical indicator systems are immobilized in a hydrogel. The hydrogel may be inserted into an optical fiber such that light may be transmitted through the hydrogel while at least a portion of the hydrogel is in contact with blood. The hydrogel is preferably permeable to blood and analytes, specifically glucose. In certain embodiments, the optical fiber together with the hydrogel comprises a glucose sensor that is placed in a mammalian (human or animal) blood vessel.

Examples of glucose-sensing chemical indicator systems and glucose sensor configurations for intravascular glucose monitoring include the optical sensors disclosed in U.S. Pat. Nos. 5,137,033, 5,512,246, 5,503,770, 6,627,177, 7,417,164 and 7,470,420, and U.S. Patent Publ. Nos. 2008/0188722, 2008/0188725, 2008/0187655, 2008/0305009, 2009/0018426, 2009/0018418, and co-pending U.S. patent application Ser. Nos. 11/296,898, 12/187,248, 12/172,059, 12/274,617 and 12/424,902; each of which is incorporated herein in its entirety by reference thereto.

Light may be transmitted into an optical glucose sensor from a light source. In certain embodiments, the light source is a light emitting diode that emits an optical excitation signal. The optical excitation signal excites the fluorophore system (s), such that the fluorophores emit light at an emission wavelength. In certain embodiments, the fluorophore systems are configured to emit an optical emission signal at a first wavelength having an intensity related to the blood glucose concentration in the blood vessel. In certain embodiments, light is directed out of the glucose sensor such that the light is detected by at least one detector. The at least one detector preferably measures the intensity of the optical emission sig-

nal, which is related to the glucose concentration present in the blood. Various optical configurations for interrogating glucose-sensing chemical indicator systems with one or more excitation light signals and for detecting one or more emission light signals from the chemical indicator systems may be employed, see e.g., U.S. patent application Ser. No. 12/027, 158 (published as 2008/0188725); incorporated herein in its entirety by reference thereto.

#### Glucose Activity and Tight Glycemic Control

While clinicians have used insulin for decades to regulate glucose levels in diabetics and critically ill patients, determining precise dosages remains a problem. Insulin reduces circulating glucose levels through a series of complex interactions involving a number of hormones and cell types. Dosage protocols for insulin attempt to replicate the physiologic secretion of the hormone by the pancreas. However, administering according to fixed times and algorithms based on blood glucose measurements can only crudely approximate the ability of a healthy individual to continuously adjust insulin production in response to the amount of bioavailable glucose and the needs of the body. Thus, to determine the precise amount of insulin that should be administered to maintain a patient's blood glucose at an appropriate level, it is necessary to have near real-time, accurate measurements of the amount of bioavailable glucose circulating in blood.

Unfortunately, most existing methods for determining blood glucose concentrations fail to provide near real-time, accurate measurements of the amount of bioavailable glucose. Clinicians and diabetic patients typically rely on point-of-care testing that seems to measure glucose concentration in plasma, e.g., using glucometers to read test strips that filter separate plasma from cells in a drop of whole blood. While the results can be available quickly, they vary depending on the patient's hematocrit, plasma protein and lipid profiles, etc., and can often be falsely elevated (See e.g., Chakravarthy et al., 2005 "Glucose determination from different vascular compartments by point-of-care testing in critically ill patients" *Chest* 128(4) October, 2005 Supplement: 220S-221S). More accurate determinations can be obtained by first separating the cellular components of whole blood. However, this requires separation of the plasma from the cellular components of blood, e.g., by centrifugation. Subsequently, isolated plasma must be stored and/or transported and/or diluted prior to analysis. Storage and processing conditions, e.g., temperature, dilution, etc., will almost certainly perturb the in vivo equilibrium between the bound and free (bioavailable) glucose. Consequently, regardless of the technology subsequently employed for measuring plasma glucose concentration (e.g., glucose oxidase, mass spectrometry, etc.), the measured glucose concentration is likely no longer reflective of the amount of bioavailable glucose in vivo. Therefore, it is not feasible to use plasma glucose measurements for near real-time monitoring and adjustment of a patient's glucose level.

Accordingly, in certain embodiments, the preferred glucose sensors described herein measure glucose "activity" as opposed to glucose concentration. More precisely, glucose activity refers to the amount of free glucose per kilogram of water. In some embodiments, glucose activity can be measured directly using glucose sensors, such as the equilibrium, non-consuming optical glucose measurement systems discussed above, which employ a chemical indicator system to quantify the amount of free, bioavailable glucose, which is in equilibrium between the water compartment of blood (i.e., not associated with cells, proteins or lipids, etc.) and the glucose binding moiety/quencher. The discussion of the sensors that follow will often refer to the physical quantity to be measured as an "analyte concentration", "glucose concentra-

tion" or simply a "concentration." However, it is to be understood that "concentration" as used herein, refers to both "analyte concentration" as that phrase would be ordinarily used and also to "activity" (in some cases "glucose activity") as that phrase is described above.

#### Optical Glucose Sensor Configurations

With reference to FIGS. 1A-D, certain prior art embodiments (see US Patent Publication No. 2008/0188725) are illustrated. The glucose sensor 117 in FIG. 1A is an optical fiber with a series of holes 116 drilled straight through the sides of the optical fiber. In certain embodiments, the holes 116 are filled with one or more glucose-sensing chemical indicator systems. These holes may be covered with a selectively permeable membrane, wherein the permeability is selected such that the molecules of the chemical indicator system (e.g., fluorophore and quencher) are retained within the cavities, whereas glucose is freely permeable. In certain embodiments, the series of holes 116 that are drilled through the glucose sensor 117 are evenly spaced horizontally and evenly rotated around the sides of the glucose sensor 117 to form a spiral or helical configuration. In certain embodiments, the series of holes are drilled through the diameter of the glucose sensor.

With reference to FIG. 1B, in certain embodiments, the glucose sensor 117 is a solid optical fiber with a series of holes 116 drilled through the sides of the fiber at an angle. In certain embodiments, the series of holes drilled at an angle, which are filled with hydrogel/chemical indicator system, are evenly spaced horizontally and evenly rotated around the sides of the glucose sensor 117. With reference to FIG. 1C, in certain embodiments, the optical fiber comprises a groove 116 along the length of the optical fiber, wherein the groove is filled with hydrogel/chemical indicator system. In certain embodiments, the depth of the groove extends to the center of the optical fiber. In certain embodiments, the groove spirals around the optical fiber. In certain embodiments, the groove spirals around the optical fiber to complete at least one rotation. In certain embodiments, the groove spirals around the optical fiber to complete multiple rotations around the optical fiber.

With reference to FIG. 1D, in certain embodiments, the glucose sensor 117 is a solid optical fiber with triangular wedges 116 cut from the fiber. In certain embodiments, the triangular wedge areas are filled with hydrogel/chemical indicator system. In certain embodiments, the triangular wedges cut-outs are evenly spaced horizontally and around the sides of the glucose sensor 117. In certain embodiments, all light traveling in the glucose sensor 117 is transmitted through at least one hole or groove 116 filled with hydrogel/chemical indicator system.

In certain embodiments, as illustrated in FIGS. 2-6, the glucose sensor 117 comprises an optical fiber 130 having a distal end 132, an atraumatic tip portion 134 having a proximal end 136 and a distal end 138, a void or cavity 116 between the distal end 132 of the optical fiber 130 and the proximal end 136 of the atraumatic tip portion 134, and a rod 140 connecting the distal end 132 of the optical fiber 130 to the proximal end 136 of the atraumatic tip portion 134, wherein the rod traverses the void or cavity. In preferred embodiments, molecules of a chemical indicator system are disposed within the void or cavity 116 and immobilized (by covalent bonding or non-covalent interaction) or otherwise associated within hydrogel matrices. See e.g., the chemical indicator systems disclosed in U.S. Pat. Nos. 7,417,164 and 7,470,420. The cavity 116 may be loaded with hydrogel/chemical indicator system by any methods known in the art. In preferred embodiments, the cavity 116 is filled with hydrogel/chemical indicator system in a liquid state. The hydrogel/chemical indica-

tor systems are preferably polymerized in situ, as detailed in co-pending U.S. patent application Ser. No. 12/026,396 (published as 2008/0187655).

In certain embodiments, the rod **140** is attached to the optical fiber **130** and/or atraumatic tip **134** by heating and expanding the optical fiber **130** and atraumatic tip **134** and embedding the rod **140** there between. In certain embodiments, the optical fiber **130** is heated to between about 100° C. and about 160° C., more preferably between about 110° C. and about 140° C. In other embodiments, the optical fiber **130** is first heated and then cooled one or more times. In certain embodiments, the rod **140** is attached to the optical fiber **130** and/or the atraumatic tip **134** by applying an adhesive. In preferred embodiments, the adhesive is biocompatible, such as for example, cyanoacrylates, epoxies, light cure adhesives, silicones, and urethanes. In certain embodiments, after applying the adhesive and joining the rod **140** with the optical fiber **130** and atraumatic tip **134**, the adhesive is cured at room temperature, by heating, or by applying UV/visible light. In certain embodiments, the time to fix the rod **140** to the optical fiber **130** and/or atraumatic tip **134** can vary from about 5 seconds to about 60 seconds, from about 15 minutes to about 5 hours, from about 60 seconds to about 10 minutes, or up to about 24 hours.

In some embodiments, the proximal surface of the rod **144** is reflective so that a portion of the excitation light signal (or reference light signal) is reflected proximally down the optical fiber **130** to a detector (not shown). The term rod is used herein to refer to any elongate structural member, regardless of its geometry, configured to connect the atraumatic tip portion to the optical fiber. The rod may be centered coaxially (as illustrated) or off-centered with regard to the cross-section of the fiber and atraumatic tip portion. In some embodiments, there may be more than one rod extending between the fiber and the atraumatic tip portion. Where more than one rod is employed, the rods may be arranged symmetrically or asymmetrically with respect to the cross-section of the fiber and atraumatic tip portion.

In certain embodiments, as illustrated in FIGS. **5** and **6**, a reference material **190** may be attached to the proximal surface of the rod **144**. The reference material **190** may be reflective (e.g., a mirror) and functions similar to embodiments in which the proximal surface of the rod **144** reflects at least a portion of the excitation light signal (or reference light signal) down the optical fiber **130** to a detector (not shown). In other embodiments, the reference material **190** comprises a separate dye indicator system, such as for example a glucose-insensitive fluorescent dye. The excitation light from the optical fiber **130** causes the glucose-insensitive fluorophore to emit a fluorescent light back to a detector (not shown) in order to reference the emission signal from the hydrogel/chemical indicator system. In certain embodiments, the separate dye indicator system is formed of a plastic material, such as for example polycarbonate, polyethylene, or polystyrene, infused with a fluorescent dye configured to emit a separate glucose-insensitive signal.

The hydrogel and glucose-sensing chemical indicator system is disposed within the cavity **116**. In preferred embodiments, the hydrogel/chemical indicator system filled cavity **116** is covered by a selectively permeable membrane **142** that allows passage of glucose into and out of the hydrogel/chemical indicator system. Although these embodiments are described using a glucose sensor **117**, it should be understood by a person of ordinary skill in the art that the sensor **117** can be modified to measure other analytes by changing, for example, the sensing chemistry, and if necessary, the selectively permeable membrane **142**.

In certain embodiments, the selectively permeable membrane **142** is attached to the optical fiber **130** and the atraumatic tip **134** by means of an adhesive. In preferred embodiments, the adhesive is biocompatible, such as for example, cyanoacrylates, epoxies, light cure adhesives, silicones, and urethanes. In certain embodiments, after applying the adhesive and attaching the selectively permeable membrane **142** to the optical fiber **130** and atraumatic tip **134**, the adhesive is cured at room temperature, by heating, or by applying UV/visible light. In certain embodiments, the time to adhere the selectively permeable membrane **142** to the optical fiber **130** and/or atraumatic tip **134** can vary from about 5 seconds to about 60 seconds, from about 15 minutes to about 5 hours, from about 60 seconds to about 10 minutes, or up to about 24 hours. In other embodiments, the selectively permeable membrane **142** is pre-fabricated as a sleeve. The sleeve may be slid into place and sealed using an adhesive and/or heated to form-fit the glucose sensor **117**. In certain embodiments, the selectively permeable membrane **142** surrounds the entire circumference of the glucose sensor **117**. In other embodiments, the selectively permeable membrane **142** covers a window **180** or opening in the glucose sensor **117** exposing the void or cavity **116** to analytes in the blood stream.

In some embodiments, as illustrated in FIGS. **2A** and **2B**, the sensor **117** comprises a distal portion and a proximal portion. The distal portion of the sensor **117** comprises the atraumatic tip portion **134**, the hydrogel/chemical indicator system filled cavity **116**, the rod **140**, at least the portion of the selectively permeable membrane **142** that covers the cavity **116** and the distal end **132** of the optical fiber **130**. The proximal portion of the sensor **117** comprises the proximal portion of the optical fiber **130**. In some embodiments, the diameter, **D1**, of the distal portion of the sensor **117** is greater than the diameter, **D2**, of the proximal portion of the sensor **117**. For example, the diameter **D1** of the distal portion of the sensor **117** can be between about 0.0080 inches and 0.020 inches, while the diameter **D2** of the proximal portion of the sensor **117** can be between about 0.005 inches to 0.015 inches. In some embodiments, the diameter **D1** of the distal portion of the sensor **117** is about 0.012 inches, while the diameter **D2** of the proximal portion of the sensor **117** is about 0.010 inches.

In some embodiments, the sensor **117**, including the selectively permeable membrane **142**, has a smooth surface. The smooth surface can be made for example by the method disclosed in co-pending U.S. patent application Ser. No. 12/026,396 (published as 2008/0187655). In summary, one preferred embodiment of the method comprises filling the cavity **116** with a solution comprising a monomer, crosslinker and an initiator, such as a thermal initiator. After the sensor **117** has been filled, the sensor **117** is dipped into liquid wax, which is allowed to harden around the sensor **117** and selectively permeable membrane **142**.

The liquid wax has a melting point that is greater than the thermal initiation temperature. Therefore, in order to reduce the likelihood of initiation during the wax dipping and coating step, the filled sensor **117** can be chilled before the wax dipping and coating step. After the sensor **117** has been coated with wax, the solution in the cavity **116** can be deoxygenated by placing the sensor **117** in a water bath while bubbling an inert gas, such as nitrogen, in the bath.

After deoxygenation, polymerization can be initiated by heating the sensor **117** to a temperature above the thermal initiation temperature, but below the melting point of the wax. Once the solution is substantially polymerized into the hydro-

gel, the wax can be removed from the sensor by use of a solvent, such as hexane, leaving a sensor 117 with a smooth surface.

In some embodiments, as illustrated in FIGS. 3A and 3B, the sensor 117 comprises a distal portion and a proximal portion with substantially the same diameter. In some embodiments, the diameter of the sensor 117 is between about 0.005 inches and 0.020 inches. In other embodiments, the diameter of the sensor 117 is between about 0.008 inches and 0.014 inches. In other embodiments, the diameter of the sensor 117 is about 0.010 inches or about 0.012 inches.

In some embodiments, the rod 140 has a proximal portion that is connected to the distal portion of the optical fiber 130 and a distal portion that is connected to the proximal portion of the atraumatic tip portion 134. The rod 140 can be made of a metal, metal alloy, plastic, polymer, ceramic, composite material or any other material with suitable mechanical properties for connecting the atraumatic tip portion 134 with the distal portion of the optical fiber 130. For example, the rod 140 can be made of stainless steel, titanium or Nitinol. The rod 140 can be cylindrical or noncylindrical, such as a bar with a square, rectangular, oval or oblong cross-section. In some embodiments, the diameter of the rod 140 is generally between about 0.001 to 0.010 inches. In other embodiments, the diameter of the rod 140 is generally between about 0.004 to 0.008 inches. In other embodiments, the diameter of the rod 140 is about 0.001 inches, about 0.002 inches or 0.004 inches. In some embodiments, the diameter of the rod 140 may be less than about 0.001 inches. In some embodiments, the diameter of the rod 140 may be greater than about 0.010 inches. In some embodiments, the length of the rod 140 is generally less than about 0.005 inches. In some embodiments, the length of the rod 140 is between about 0.005 to 0.040 inches. In other embodiments, the length of the rod 140 is generally between about 0.020 to 0.040 inches. In other embodiments, the length of the rod 140 is generally about 0.015 inches. In some embodiments, the length of the rod 140 is generally greater than about 0.005 inches.

The rod 140 adds mechanical stability to the distal portion of the sensor 117. In some embodiments, the rod 140 also adds flexibility to the distal portion of the sensor 117, allowing the atraumatic tip portion 134 to flex back and forth relative to the orientation of the optical fiber 130. The flexibility of the rod 140, and thus the degree which the atraumatic tip portion 134 can flex, can be increased or decreased by decreasing or increasing the diameter of the rod 140. In addition, the flexibility of the rod 140 can be altered by making the rod 140 from a stiff or flexible material.

In some embodiments as illustrated in FIGS. 2A and 3A, a reflective surface 144 is disposed on the proximal end of the rod 140, which is inserted into the optical fiber 130. The reflective surface 144 is capable of reflecting back at least a portion of either reference light or excitation light emitted from the light source. The other end of the rod 140 is inserted into the atraumatic tip portion 134. In certain embodiments, the atraumatic tip portion may be made from a non-reflective material, for example polyethylene (e.g., black polyethylene) or polypropylene. The reference or excitation light that passes through the optical fiber 130 in the region corresponding to the diameter of the rod 140 is reflected off the reflective surface 144 without entering into the hydrogel filled cavity 116; the amount of light entering the hydrogel/chemical indicator system can be controlled by varying the diameter/cross-sectional area of the rod and/or by attaching a mirror or other reflective member 190 (illustrated in FIGS. 5 and 6) having a selected cross-sectional area to the proximal end of the rod. The hydrogel filled cavity 116 is preferably covered by a selectively

permeable membrane 142, which is at least permeable to glucose. Therefore, the reflected reference or excitation light and the ratio between the reflected and emitted light is independent of the temperature, pH, glucose concentration, and chemistry formulation of the hydrogel filled cavity 116. The ratio between the reflected and emitted light is dependent, however, on the diameter of the rod and the ratio of the diameter of the rod to the area of the sensor. In certain embodiments, the rod 140 is sufficiently stiff to keep the hydrogel filled cavity 116 in a fixed orientation relative to the optical fiber 130 so that any light that is transmitted through the hydrogel cavity 116 is not reflected back to the optical fiber 130.

With regard to FIG. 3B, there is shown a perspective view of the distal region of the sensor 117 illustrated in FIG. 3A. It can be appreciated in the illustrated embodiment that there is no change in sensor diameter from the optical fiber 130, through the membrane 142 covered hydrogel cavity, until the tapered atraumatic distal tip portion 134.

In some embodiments, as shown in FIG. 4, a window 180 is cut into a hypotube 140. The distal end 132 of the optical fiber 130 is inserted onto the reflective surface 144 (e.g., an annular mirror) and then heated such that the optical fiber 130 swells to fully contact the reflective surface 144 of the hypotube 140. In certain embodiments, heating the optical fiber 130 is carried out in a glass tube. In other embodiments, heating the optical fiber 130 is carried out in an oven. In still other embodiments, the optical fiber 130 is attached to the hypotube 140 using an adhesive. In preferred embodiments, the adhesive is biocompatible, such as for example, cyanoacrylates, epoxies, light cure adhesives, silicones, and urethanes. In certain embodiments, after applying the adhesive and attaching the optical fiber 130 to the hypotube 140, the adhesive is cured at room temperature, by heating, or by applying UV/visible light. In certain embodiments, the time to adhere the optical fiber 130 to the hypotube 140 can vary from about 5 seconds to about 60 seconds, from about 15 minutes to about 5 hours, from about 60 seconds to about 10 minutes, or up to about 24 hours. Similar methods may be employed for attaching the hypotube 140 to the atraumatic tip 134. Similar to previous embodiments, the reference or excitation light is reflected off the reflective surface 144 without entering the window 180 that opens to the cavity 116 which is filled with hydrogel/chemical indicator system. Therefore, the reflected reference or excitation light and the ratio between the reflected and emitted lights is independent of the temperature, pH, glucose concentration, and hydrogel chemistry. The surface area of the reflective surface can be varied to control the amount of excitation light that enters the hydrogel/chemical indicator system filled cavity 116. The distal end 136 of the hypotube 140, as in previous embodiments, may have a non-reflective surface, such as a black plug made of polyethylene so that light entering the hydrogel/chemical indicator system filled cavity 116 is not reflected back into the optical fiber 130. In some embodiments, the window 180 containing the hydrogel/chemical indicator system filled cavity 116 is covered by a selectively permeable membrane (not shown).

In some embodiments, as illustrated in FIG. 5, the glucose sensor 117 includes a cage 195, as an outer shell, connecting the atraumatic tip 134 with the optical fiber 130. The cage 195 can add mechanical stability to the distal portion of the sensor 117. In some embodiments, the cage 195 also adds flexibility to the distal portion of the sensor 117, allowing the atraumatic tip portion 134 to flex back and forth relative to the orientation of the optical fiber 130. The flexibility of the cage 195, and thus the degree which the atraumatic tip portion 134 can flex,

can be increased or decreased by decreasing or increasing the thickness of the cage 195 walls. In addition, the flexibility of the cage 195 can be altered by making the cage 195 from a stiff or flexible material. In certain embodiments, the thickness of the cage 195 walls is about 0.001 inches, about 0.002 inches, or about 0.004 inches. In some embodiments, the thickness of the cage 195 walls may be less than about 0.001 inches. In some embodiments, the thickness of the cage 195 walls may be greater than about 0.010 inches.

In some embodiments, the diameter of the optical fiber 130 may be smaller than the diameter of the interior of the cage 195, allowing the optical fiber 130 to fit within the interior of the cage 195 and abut the void or cavity 116. For example, the diameter of the optical fiber 130 may be between about 0.005 inches and about 0.020 inches, between about 0.008 inches and about 0.014 inches, or between about 0.010 inches and about 0.012 inches. The diameter of the interior of the cage 195 may be about 0.001 inches larger.

In certain embodiments, the cage 195 has a window or opening 180, covered by a selectively permeable membrane 142 (not shown), which allows for at least the transmission of analytes, such as a glucose, into the void or cavity 116. In certain preferred embodiments, the void or cavity 116 is filled with a hydrogel/chemical indicator system. A rod 140 may be positioned within the glucose sensor 117 having a reference material 190. As discussed above, the reference material 190 may be a mirror for reflecting excitation light from the optical fiber 130 back to a detector (not shown) or a glucose-insensitive fluorescent dye for emitting a glucose-insensitive reference signal back to a detector (not shown). The combination of the cage 195 and the rod 140 may provide a sufficiently rigid structure such that the excitation light which enters the void or cavity 116 remains separate from the light that enters the reference material 190.

In some embodiments, as illustrated in FIG. 6, the glucose sensor 117 does not have a cage 195 surrounding the void or cavity 116. Instead, similar to FIGS. 2A and 3A, the rod 140 connects the optical fiber 130 and atraumatic tip 134, providing structure for the glucose sensor 117, and is surrounded by the void or cavity 116, which in turn is covered by a selectively permeable membrane 142. Similar to FIG. 3A, the diameter of the optical fiber 130 is the same as the diameter of the hydrogel/chemical indicator system encased cavity 116. As discussed with respect to FIG. 5, the rod may have a reference material 190 attached to the proximal surface of the rod 144, which functions as previously discussed.

FIGS. 7-11 illustrate certain embodiments having different configurations for the reference material 190. As discussed previously, the reference material 190 in each of these embodiments may either comprise reflective material to return at least a portion of the excitation light back to a detector (not shown) or a separate dye indicator system to return an emission signal back to a detector (not shown). Similar to previous embodiments, the excitation or reference light is reflected off the reflective surface 190 without entering the cavity 116, which is filled with the hydrogel/chemical indicator system. Likewise, the emitted or reference light from the separate dye indicator system is independent of the glucose concentration. Therefore, the reference light and the ratio between the reference light and emitted glucose concentration dependent lights are independent of the temperature, pH, glucose concentration, and hydrogel chemistry. The surface area, shape, and configuration of the reference material 190 can be varied to control the amount of excitation light that enters the hydrogel/chemical indicator system filled cavity 116. The distal end 136 of the rod or hypotube 140, as in previous embodiments, or reference material 190 may have a

non-reflective surface, such as a black plug made of polyethylene so that light entering the hydrogel/chemical indicator system filled cavity 116 is not reflected back into the optical fiber 130.

In FIG. 7, the reference material 190 abuts the void or cavity 116 beneath the cage 195 and extends to and comprises the atraumatic tip 134. In certain embodiments, the atraumatic tip 134 is formed of a glucose-insensitive red dye plastic material. In FIG. 8, the reference material 190 is a reflective strip that spans the diameter of the hydrogel-filled cavity 116. The term reflective strip is used herein to refer to any elongate member, regardless of its geometry, width, or thickness that spans at least a cross-section of the glucose sensor 117. The reflective strip 190 may be centered at the diameter of the glucose sensor 117 (as illustrated) or off-centered with regard to the cross-section of the cage 195 or optical fiber 130. In some embodiments, there may be more than one reflective strip in one or more locations within the glucose sensor 117. Where more than one reflective strip is employed, the reflective strip may be arranged symmetrically or asymmetrically with respect to the cross-section of the glucose sensor 117. In certain embodiments, the reflective strip 190 may be between about 0.001 inches and about 0.005 inches wide and between about 0.001 inches and about 0.005 inches thick.

In FIG. 9, the reference material 190 is disposed within the cage 195 as a hypotube containing the hydrogel-filled cavity 116 and having a reflective surface or annular mirror at the proximal surface of the hypotube 144. In certain embodiments, the hypotube 144 has an outer diameter equal to the outer diameter of the optical fiber 130. In FIG. 10, similar to FIG. 8, the reference material 190 is reflective strip, but the reflective strip in FIG. 10 is placed within a hole drilled in the optical fiber 130, rather than the abutting the hydrogel-filled cavity 116. FIG. 11 illustrates an embodiment in which the reference material 190 is located in the cavity 116 and is in a side-by-side configuration with the hydrogel/chemical indicator system.

In certain embodiments, as illustrated in FIG. 12, a reference material 200 comprises a translucent material. In certain embodiments, this translucent material comprises a red dye, such as the glucose-insensitive fluorescent dye discussed previously. The red dye may allow some of the excitation light to be transmitted to the hydrogel-filled cavity 116, may reflect some of the excitation light back to a detector (not shown) before the excitation light reaches the hydrogel-filled cavity 116, and may emit a separate glucose-insensitive signal to a detector (not shown).

A person skilled in the art would readily understand that the above described embodiments, or components of the above described embodiments, may be combined within the scope of the present invention. For example, a glucose sensor may contain one or more structural elements, such as a cage, a hypotube, and/or a rod within the scope of the present invention. In addition, a glucose sensor may contain one or more reference materials, functioning as a reflective surface and/or as a separate dye indicator system, in different locations and configurations within the scope of the present invention.

In some embodiments (see e.g., FIGS. 2-12), the glucose sensor 117 comprises an atraumatic tip portion 134. The atraumatic tip portion 134 has a distal end 138 that is curved and substantially free of sharp edges. In addition, the atraumatic tip portion 134 can be flexible and deformable. The distal end 138 of the atraumatic tip portion 134 can be hemispherical, parabolic, elliptical or curved in any other suitable shape that reduces the risk of injury to the patient. The

atraumatic tip portion **134** can be made from a variety of materials, such as plastics, polymers, gels, metals and composites of the above.

In some embodiments, the glucose sensor **117** includes a temperature sensor or probe **146**, such as thermocouple or thermistor (See e.g., FIG. 2A). The temperature sensor **146** can measure the temperature of the hydrogel and glucose sensing chemistry system, and/or the blood when disposed intravascularly. The temperature sensor **146** is particularly preferred when the glucose-sensing chemistry is affected by temperature. For example, in some embodiments, the fluorescence intensity emitted by the fluorophore system is dependent on the temperature of the fluorophore system. By measuring the temperature of the fluorophore system, temperature induced variations in fluorophore fluorescence intensity can be accounted for, allowing for more accurate determination of glucose concentration.

In certain embodiments, the temperature sensor can be a thermistor (as described above with regard to FIG. 2, reference numeral **146**, a platinum resistance temperature device (“RTD”), another RTD, a thermocouple, an infrared-based temperature detector, a fluorescence-based temperature sensing element, or other temperature sensing elements with determinable temperature-dependent characteristics.

Devices such as thermistors, platinum RTDs, and other RTDs generally require one or more conductors, such as wires, to conduct the output of the sensor to a receiving unit which converts the output to a temperature signal. The conductors can be bundled with the optical fiber of fluorescence-based glucose sensors, such as those discussed above, or they can be routed separately. In one embodiment, the temperature sensor is placed inside the body, and the receiver is placed outside the body. In another embodiment, the temperature sensor is placed inside the body, and a transmitter, signal processor, etc. is also placed inside the body and is connected to or is a part of the temperature sensor. In preferred embodiments, the temperature sensing element is located at or near the glucose sensing moiety.

In another embodiment, a fluorescence-based temperature sensing technique can be used. Fluorescence-based temperature sensing techniques include those based on fluorescence decay, such as where an excitation light is provided to a phosphor, the excitation light is stopped, and the fluorescence is monitored versus time, with the rate of decrease in fluorescence being related to the temperature of the phosphor. Various techniques, can also include phase measurement and phase angle analysis.

Methods for performing fluorescence-based temperature measurement have been described. See for example, LumaSense Technologies, Inc. (Santa Clara, Calif.), “Fluoroptic Temperature Monitoring,” [http://www.lumasenseinc.com/technology/fluoroptic\\_thermometry.html](http://www.lumasenseinc.com/technology/fluoroptic_thermometry.html). Fluorescent materials that can be used in fluorescence-based temperature measurement are known to, or readily identified by those having skill in the art.

In some embodiments, the fluorescent material can be surrounded by material which prevents or inhibits chemical interaction between the fluorescent material and blood components. Suitable materials include glass (for example, borosilicate, lime-soda, or other types including those used for fiberoptic cables), polymers (for example, Teflon, fluoropolymers, silicone, latex, polyolefins, polyisoprene, and other rigid and nonrigid polymeric materials), metals (for example, 300 series stainless steel, 400 series stainless steel, nickel, nickel alloys, chromium steels, zirconium and its alloys, titanium and its alloys, as well as other corrosion resistant metals and alloys including exotic metals and alloys), ceramics (for

example, ceramic materials related to aluminum oxide, silica and oxide, zirconium, carbides, etc.), and combinations of these.

In some embodiments, the temperature sensor can be positioned within the glucose sensor, or near it. While in one preferred embodiment, the temperature sensor can be positioned as close as possible to (e.g., within) the glucose-sensing chemical indicator system of the glucose sensor, positions some distance away can also be successfully utilized, including those locations where the temperature measured provides an indication of the temperature at the glucose-sensing site(s) within an acceptable error for the use for which the temperature measurement is being made.

In some embodiments, the temperature sensor and/or the leads to the sensor can be isolated from the physiological environment, such as by coating, covering, or encasing the various parts with a material that prevents or inhibits chemical or physical interaction between the temperature sensor and/or its leads and blood components. Chemical interactions that are preferably avoided include corrosion, leaching of chemical species, generation of additional signals (e.g. optical, electrical, etc.) and take-up by the body of materials present in the sensor or leads, whether present from manufacture, corrosion or other means, such as compounds, metals, or ions causing a physiological response in some patients including copper, silver, organic compounds, organometallic compounds, etc.

Physical interactions can include breakage and physical separation (e.g. disconnection and potential loss), signal leakage (e.g. optical; electrical, etc.), signal degradation (including resistance, stray signal detection, noise, capacitance, electrochemical effects, induced voltages, ground loops, etc.). Suitable materials include glass (e.g., borosilicate, lime-soda, as well as other types of glass, such as those used in production of optic fibers), polymers (e.g., Teflon, fluoropolymers, silicone, latex, polyolefins, polyisoprene, acrylics, polycarbonates, and other rigid and nonrigid polymeric materials), metals (e.g., 300 series stainless steel, 400 series stainless steel, nickel, nickel alloys, chromium steels, zirconium and its alloys, titanium and its alloys, as well as other corrosion resistant metals and alloys including exotic metals and alloys), ceramics (e.g., ceramic materials related to aluminum oxide, silica and oxide, zirconium, carbides, etc.), and combinations of these.

Suitable methods for applying for isolating material to the temperature sensor or leads can include any appropriate method, including casting, painting, dipping, gluing, reacting, drawing, depositing, mechanically adhering, encapsulating, etc.

In some embodiments, suitable sizes for temperature sensors that will be incorporated into the glucose sensor include those temperature sensing elements resulting in an overall glucose sensor of between about 0.005 inches and 0.020 inches.

FIG. 13 illustrates another embodiment for measuring the glucose concentration in comparison to a reference signal. In this embodiment, a LED source **1300** sends an excitation signal down two separate adjacent optical fibers **1310**, **1320**. The first optical fiber, or the glucose fiber **1310**, has a proximal tip and a distal tip. The distal tip has a glucose sensing hydrogel **1330** which contains a fluorophore or dye, a quencher, and glucose binding receptors. The second optical fiber, or the reference fiber **1320**, also has a proximal tip and a distal tip. The distal tip of the reference fiber has a reference material **1340**. In certain embodiments, the reference material **1340** contains the same or a different fluorophore or dye, may or may not contain the quencher, but does not contain glucose

receptors. In other embodiments, the reference material **1340** has the same exact glucose sensing hydrogel, but it is encased in a glucose impermeable membrane. In both of these embodiments, the reference fiber **1320** emits a fluorescent return signal independent of the glucose concentration.

After the excitation light passes through the glucose fiber **1310** and the reference fiber **1320**, the glucose sensing hydrogel **1330** and the reference material **1340** emit fluorescent signals back to two separate detectors, a glucose signal detector **1350** and a reference signal detector **1360**, for ratiometric processing. The benefit of the dual fiber configuration is that both fibers **1310**, **1320** experience the same external pressure, bending, temperature, and other external factors. In addition, both fibers **1310**, **1320** contain substantially the same material in the glucose sensing hydrogel **1330** and reference material **1340**. As a result, the ratio of the intensities between the two fibers **1310**, **1320**, as measured by the detectors **1350**, **1360**, produce a calibrated glucose signal that removes, inter alia, the effect of the fluctuations in the LED output or altered transmission along the optical fiber, and thereby increase the accuracy in the measurement of the glucose concentration.

With reference to FIG. **14**, in certain embodiments, the light generated by the single light source **401** is transmitted through an optical module comprising a collimator lens **402**, an interference filter **403**, and/or a focusing lens **404** as described above. The resulting light can be filtered through an interference filter **403**. The resulting light can be focused by a focusing lens **404** into an optical fiber **405**, which may be a single fiber or a bundle of fibers. The optical fiber **405** can surround optical fiber **410** as both fiber optic lines connect to the first end of the glucose sensor **407**. In certain embodiments, a mirror or reflective surface **409** is attached to the second end of the glucose sensor **407**. The optical fiber **410** may be a single fiber or a bundle of fibers. The glucose sensor can comprise hydrogels that further comprise a fluorophore system that produces two emission wavelengths, a first emission wavelength and a second emission wavelength. In certain embodiments, the fluorophore system is excited by the light generated by light source **401**. In certain embodiments, the optical fiber **410** is connected to a light sensitive module comprising a microspectrometer **411** that measures the entire spectrum of light in the glucose measurement system **400**. Data from the microspectrometer **411** can be transmitted to computer **412** for processing. The microspectrometer **411** can allow system **400** to simultaneously measure the excitation light intensity as well as both emission light intensities. Ratiometric calculations may be employed to substantially eliminate or reduce non-glucose related factors affecting the intensity of the measured emission light and measured excitation light (as detailed in US Patent Publication No. 2008/0188725; incorporated herein in its entirety by reference thereto). The measured emission light can be divided by the measured excitation light, wherein such calculations substantially eliminate or reduce non-glucose related factors affecting the intensity of the lights.

In certain preferred embodiments, the fluorophore dye may be selected such that it exists in distinguishable acid and base conformations, each of which emit at a distinct wavelength, and wherein the relative proportion of acid and base forms depend on the pH. The ratio of intensities of the acid and base emissions can be used to determine the pH of the blood (as detailed in US Patent Publication No. 2008/0188722; incorporated herein in its entirety by reference thereto). The ratio of the acid or base emission intensity over the excitation light can be used to determine the level of glucose in the blood. Of course in a variation to this single exciter-dual emitter fluorophore system, one could employ a single exciter-single

emitter for detection of glucose concentration without simultaneous ratiometric determination of pH. Indeed, a great variety of design options are available (see e.g., US Patent Publication Nos. 2008/0188725 and 2008/0188722), wherein the chemical indicator and optical systems may be selected based on the preferred use.

#### Glucose-Sensing Chemical Indicator Systems

In certain embodiments, the hydrogels are associated with a plurality of fluorophore systems. In certain embodiments, the fluorophore systems comprise a quencher with a glucose receptor site. In certain embodiments, when there is no glucose present to bind with the glucose receptor, the quencher prevents the fluorophore system from emitting light when the dye is excited by an excitation light. In certain embodiments, when there is glucose present to bind with the glucose receptor, the quencher allows the fluorophore system to emit light when the dye is excited by an excitation light.

In certain embodiments, the emission produced by the fluorophore system may vary with the pH (as well as the temperature) of the solution (for example, blood), such that different excitation wavelengths (one exciting the acid form of the fluorophore and the other the base form of the fluorophore) produce different emissions signals. In preferred embodiments, the ratio of the emission signal from the acid form of the fluorophore over the emission signal from the base form of the fluorophore is related to the pH level of the blood. In certain embodiments, an interference filter is employed to ensure that the two excitation lights are exciting only one form (the acid form or the base form) of the fluorophore. Chemical indicator systems, hardware configurations and methods for determining both pH and glucose based on ratiometric determination are described in detail in co-pending U.S. application Ser. No. 11/671,880 (published as 2008/0188722) and Ser. No. 12/027,158 (published as 2008/0188725); incorporated herein in their entirety by reference thereto.

The indicator system (also referred to herein as a fluorophore system) can comprise a fluorophore operably coupled to a quencher. In certain embodiments, the fluorophore system comprises a polymer matrix comprising a fluorophore susceptible to quenching by a viologen, a viologen quencher with quenching efficacy dependent on glucose concentration, and a glucose permeable polymer, wherein said matrix is in contact with blood in vivo. Preferably the fluorophore is a fluorescent organic dye, the quencher is a boronic acid functionalized viologen, and the matrix is a hydrogel.

“Fluorophore” refers to a substance that when illuminated by light at a particular wavelength emits light at a longer wavelength; i.e. it fluoresces. Fluorophores include but are not limited to organic dyes, organometallic compounds, metal chelates, fluorescent conjugated polymers, quantum dots or nanoparticles and combinations of the above. Fluorophores may be discrete moieties or substituents attached to a polymer.

Fluorophores that may be used in preferred embodiments are capable of being excited by light of wavelength at or greater than about 400 nm, with a Stokes shift large enough that the excitation and emission wavelengths are separable by at least 10 nm. In some embodiments, the separation between the excitation and emission wavelengths may be equal to or greater than about 30 nm. These fluorophores are preferably susceptible to quenching by electron acceptor molecules, such as viologens, and are resistant to photo-bleaching. They are also preferably stable against photo-oxidation, hydrolysis and biodegradation.

In some embodiments, the fluorophore may be a discrete compound.



compounds. The conjugated heterocyclic aromatic di-nitrogen compounds are selected from dipyriddyls, dipyridyl ethylenes, dipyridyl phenylenes, phenanthrolines, and diazafluorenes, wherein the nitrogen atoms are in a different aromatic ring and are able to form an onium salt. It is understood that all isomers of said conjugated heterocyclic aromatic di-nitrogen compounds in which both nitrogens can be substituted are useful in this invention. In one embodiment, the quencher may be one of the bis-onium salts derived from 3,3'-dipyridyl, 4,4'-dipyridyl and 4,7-phenanthroline.

In some embodiments, the viologen-boronic acid adduct may be a discrete compound having a molecular weight of about 400 daltons or greater. In other embodiments, it may also be a pendant group or a chain unit of a water-soluble or water-dispersible polymer with a molecular weight greater than about 10,000 daltons. In one embodiment, the quencher-polymer unit may be non-covalently associated with a polymer matrix and is physically immobilized therein. In yet another embodiment, the quencher-polymer unit may be immobilized as a complex with a negatively charge water-soluble polymer.

In other embodiments, the viologen-boronic acid moiety may be a pendant group or a chain unit in a crosslinked, hydrophilic polymer or hydrogel sufficiently permeable to the analyte (e.g., glucose) to allow equilibrium to be established.

In other embodiments, the quencher may be covalently bonded to a second water-insoluble polymer matrix  $M^2$ , which can be represented by the structure  $M^2-L^2-Q$ .  $L^2$  is a linker selected from the group consisting of a lower alkylene (e.g.,  $C_1$ - $C_8$  alkylene), sulfonamide, amide, quaternary ammonium, pyridinium, ester, ether, sulfide, sulfone, phenylene, urea, thiourea, urethane, amine, and a combination thereof. The quencher may be linked to  $M^2$  at one or two sites in some embodiments.

In certain embodiments, at least one quencher precursor is used to attach the quenching moiety to at least one polymer. For example, aromatic groups may be used to functionalize a viologen with combinations of boronic acid groups and reactive groups. In certain embodiments, this process includes attaching an aromatic group to each of the two nitrogens in the dipyridyl core of the viologen. At least one boronic acid group, a reactive group, or a combination of the two are then attached to each aromatic group, such that the groups attached to each of the two nitrogens on the dipyridyl core of the viologen may either be the same or different. Certain combinations of the functionalized viologen quenching moiety are described as follows:

a) a first aromatic group having a pendent reactive group is attached to the first nitrogen and a second aromatic group having at least one pendent boronic group is attached to the second nitrogen;

b) one or more boronic acid groups are attached to a first aromatic group, which is attached to the first nitrogen, and one boronic acid group and a reactive group are attached to a second aromatic group, which second aromatic group is attached to the second nitrogen;

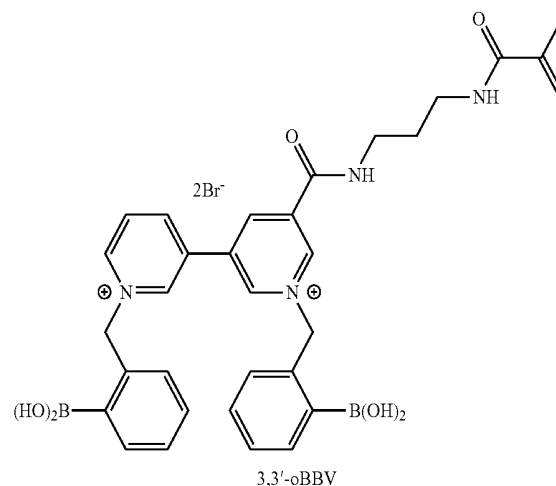
c) one boronic acid group and a reactive group are attached to a first aromatic group, which first aromatic group is attached to the first nitrogen, and one boronic acid group and a reactive group are attached to a second aromatic group, which is attached to the second nitrogen; and

d) one boronic acid group is attached to an aromatic group, which aromatic group is attached to each of the two nitrogens, and a reactive group is attached to a carbon in a heteroaromatic ring in the heteroaromatic centrally located group.

Preferred embodiments comprise two boronic acid moieties and one polymerizable group or coupling group wherein

the aromatic group is a benzyl substituent bonded to the nitrogen and the boronic acid groups are attached to the benzyl ring and may be in the ortho- meta- or para-positions.

In one preferred embodiment, the quencher precursor (before incorporation into a hydrogel) may be 3,3'-oBBV (structure illustrated below); see U.S. Pat. No. 7,470,420, incorporated in its entirety herein by reference thereto.



The quencher precursor 3,3'-oBBV may be used with HPTS-CysMA to make hydrogels in accordance with preferred aspects of the invention.

Other indicator chemistries, such as those disclosed in U.S. Pat. No. 5,176,882 to Gray et al. and U.S. Pat. No. 5,137,833 to Russell, can also be used in accordance with embodiments of the present invention; both of which are incorporated herein in their entireties by reference thereto. In some embodiments, an indicator system may comprise an analyte binding protein operably coupled to a fluorophore, such as the indicator systems and glucose binding proteins disclosed in U.S. Pat. Nos. 6,197,534, 6,227,627, 6,521,447, 6,855,556, 7,064,103, 7,316,909, 7,326,538, 7,345,160, and 7,496,392, U.S. Patent Application Publication Nos. 2003/0232383, 2005/0059097, 2005/0282225, 2009/0104714, 2008/0311675, 2008/0261255, 2007/0136825, 2007/0207498, and 2009/0048430, and PCT International Publication Nos. WO 2009/021052, WO 2009/036070, WO 2009/021026, WO 2009/021039, WO 2003/060464, and WO 2008/072338 which are hereby incorporated by reference herein in their entireties.

For in vivo applications, the sensor is used in a moving stream of physiological fluid which contains one or more polyhydroxyl organic compounds or is implanted in tissue such as muscle which contains said compounds. Therefore, it is preferred that none of the sensing moieties escape from the sensor assembly. Thus, for use in vivo, the sensing components are preferably part of an organic polymer sensing assembly. Soluble dyes and quenchers can be confined by a selectively permeable membrane that allows passage of the analyte but blocks passage of the sensing moieties. This can be realized by using as sensing moieties soluble molecules that are substantially larger than the analyte molecules (molecular weight of at least twice that of the analyte or greater than 1000 preferably greater than 5000); and employing a selective semipermeable membrane such as a dialysis or an ultrafiltration membrane with a specific molecular weight cutoff between the two so that the sensing moieties are quantitatively retained.

Preferably the sensing moieties are immobilized in an insoluble polymer matrix, which is freely permeable to glucose. The polymer matrix is comprised of organic, inorganic or combinations of polymers thereof. The matrix may be composed of biocompatible materials. Alternatively, the matrix is coated with a second biocompatible polymer that is permeable to the analytes of interest.

The function of the polymer matrix is to hold together and immobilize the fluorophore and quencher moieties while at the same time allowing contact with the analyte, and binding of the analyte to the boronic acid. To achieve this effect, the matrix must be insoluble in the medium, and in close association with it by establishing a high surface area interface between matrix and analyte solution. For example, an ultra-thin film or microporous support matrix is used. Alternatively, the matrix is swellable in the analyte solution, e.g. a hydrogel matrix is used for aqueous systems. In some instances, the sensing polymers are bonded to a surface such as the surface of a light conduit, or impregnated in a microporous membrane. In all cases, the matrix must not interfere with transport of the analyte to the binding sites so that equilibrium can be established between the two phases. Techniques for preparing ultra-thin films, microporous polymers, microporous sols, and hydrogels are established in the art. All useful matrices are defined as being analyte permeable.

Hydrogel polymers are used in some embodiments. The term, hydrogel, as used herein refers to a polymer that swells substantially, but does not dissolve in water. Such hydrogels may be linear, branched, or network polymers, or polyelectrolyte complexes, with the proviso that they contain no soluble or leachable fractions. Typically, hydrogel networks are prepared by a crosslinking step, which is performed on water-soluble polymers so that they swell but do not dissolve in aqueous media. Alternatively, the hydrogel polymers are prepared by copolymerizing a mixture of hydrophilic and crosslinking monomers to obtain a water swellable network polymer. Such polymers are formed either by addition or condensation polymerization, or by combination process. In these cases, the sensing moieties are incorporated into the polymer by copolymerization using monomeric derivatives in combination with network-forming monomers. Alternatively, reactive moieties are coupled to an already prepared matrix using a post polymerization reaction. Said sensing moieties are units in the polymer chain or pendant groups attached to the chain.

The hydrogels useful in this invention are also monolithic polymers, such as a single network to which both dye and quencher are covalently bonded, or multi-component hydrogels. Multi-component hydrogels include interpenetrating networks, polyelectrolyte complexes, and various other blends of two or more polymers to obtain a water swellable composite, which includes dispersions of a second polymer in a hydrogel matrix and alternating microlayer assemblies.

Monolithic hydrogels are typically formed by free radical copolymerization of a mixture of hydrophilic monomers, including but not limited to HEMA, PEGMA, methacrylic acid, hydroxyethyl acrylate, N-vinyl pyrrolidone, acrylamide, N,N'-dimethyl acrylamide, and the like; ionic monomers include methacryloylaminopropyl trimethylammonium chloride, diallyl dimethyl ammonium chloride, vinyl benzyl trimethyl ammonium chloride, sodium sulfopropyl methacrylate, and the like; crosslinkers include ethylene dimethacrylate, PEGDMA, trimethylolpropane triacrylate, and the like. The ratios of monomers are chosen to optimize network properties including permeability, swelling index, and gel strength using principles well established in the art. In one embodiment, the dye moiety is derived from an ethyleni-

cally unsaturated derivative of a dye molecule, such as 8-acetoxypyrene-1,3,6-N,N',N''-tris(methacrylamidopropylsulfonamide), the quencher moiety is derived from an ethylenically unsaturated viologen such as 4-N-(benzyl-3-boronic acid)-4'-N'-(benzyl-4-ethenyl)-dipyridinium dihalide (m-SBBV) and the matrix is made from HEMA and PEGDMA. The concentration of dye is chosen to optimize emission intensity. The ratio of quencher to dye is adjusted to provide sufficient quenching to produce the desired measurable signal.

In some embodiments, a monolithic hydrogel is formed by a condensation polymerization. For example, acetoxypyrene trisulfonyl chloride is reacted with an excess of PEG diamine to obtain a tris-(amino PEG) adduct dissolved in the unreacted diamine. A solution of excess trimesoyl chloride and an acid acceptor is reacted with 4-N-(benzyl-3-boronic acid)-4'-N'-(2 hydroxyethyl) bipyridinium dihalide to obtain an acid chloride functional ester of the viologen. The two reactive mixtures are brought into contact with each other and allowed to react to form the hydrogel, e.g. by casting a thin film of one mixture and dipping it into the other.

In other embodiments, multi-component hydrogels wherein the dye is incorporated in one component and the quencher in another are preferred for making the sensor of this invention. Further, these systems are optionally molecularly imprinted to enhance interaction between components and to provide selectivity for glucose over other polyhydroxy analytes. Preferably, the multicomponent system is an interpenetrating polymer network (IPN) or a semi-interpenetrating polymer network (semi-IPN).

The IPN polymers are typically made by sequential polymerization. First, a network comprising the quencher is formed. The network is then swollen with a mixture of monomers including the dye monomer and a second polymerization is carried out to obtain the IPN hydrogel.

The semi-IPN hydrogel is formed by dissolving a soluble polymer containing dye moieties in a mixture of monomers including a quencher monomer and through complex formation with the fluorophore. In some embodiments, the sensing moieties are immobilized by an insoluble polymer matrix which is freely permeable to polyhydroxyl compounds. Additional details on hydrogel systems have been disclosed in US Patent Publications Nos. US2004/0028612, and 2006/0083688 which are hereby incorporated by reference in their entireties.

The polymer matrix is comprised of organic, inorganic or combinations of polymers thereof. The matrix may be composed of biocompatible materials. Alternatively, the matrix is coated with a second biocompatible polymer that is permeable to the analytes of interest. The function of the polymer matrix is to hold together and immobilize the fluorescent dye and quencher moieties while at the same time allowing contact with the analytes (e.g., polyhydroxyl compounds, H<sup>+</sup> and OH<sup>-</sup>), and binding of the polyhydroxyl compounds to the boronic acid. Therefore, the matrix is insoluble in the medium and in close association with it by establishing a high surface area interface between matrix and analyte solution. The matrix also does not interfere with transport of the analyte to the binding sites so that equilibrium can be established between the two phases. In one embodiment, an ultra-thin film or microporous support matrix may be used. In another embodiment, the matrix that is swellable in the analyte solution (e.g. a hydrogel matrix) can be used for aqueous systems. In some embodiments, the sensing polymers are bonded to a surface such as the surface of a light conduit, or impregnated in a microporous membrane. Techniques for preparing ultra-

thin films, microporous polymers, microporous sol-gels, and hydrogels have been established in the prior art.

In one preferred embodiment, the boronic acid substituted viologen may be covalently bonded to a fluorescent dye. The adduct may be a polymerizable compound or a unit in a polymer. One such adduct for example may be prepared by first forming an unsymmetrical viologen from 4,4'-dipyridyl by attaching a benzyl-3-boronic acid group to one nitrogen and an aminoethyl group to the other nitrogen atom. The viologen is condensed sequentially first with 8-acetoxypyrene-1,3,6-trisulfonyl chloride in a 1:1 mole ratio followed by reaction with excess PEG diamine to obtain a prepolymer mixture. An acid acceptor is included in both steps to scavenge the byproduct acid. The prepolymer mixture is crosslinked by reaction with a polyisocyanate to obtain a hydrogel. The product is treated with base to remove the acetoxyl blocking group. Incomplete reaction products and unreacted starting materials are leached out of the hydrogel by exhaustive extraction with deionized water before further use. The product is responsive to glucose when used as the sensing component as described herein.

Alternatively, such adducts are ethylenically unsaturated monomer derivatives. For example, dimethyl bis-bromomethyl benzene boronate is reacted with excess 4,4'-dipyridyl to form a half viologen adduct. After removing the excess dipyrindyl, the adduct is further reacted with an excess of bromoethylamine hydrochloride to form the bis-viologen adduct. This adduct is coupled to a pyranine dye by reaction with the 8-acetoxypyrene-tris sulfonyl chloride in a 1:1 mole ratio in the presence of an acid acceptor followed by reaction with excess aminopropylmethacrylamide. Finally, any residual amino groups may be reacted with methacryloyl chloride. After purification, the dye/viologen monomer may be copolymerized with HEMA and PEGDMA to obtain a hydrogel.

#### Solution Example

To a solution of HPTS-CysMA ( $1 \times 10^{-5}$  M in pH 7.4 PBS) was added increasing amounts of 3,3'-oBBV (30 mM in MeOH) and the fluorescence emission measured after each addition. FIG. 15 gives the relative emission change (Stern-Volmer curve) upon addition of 3,3'-oBBV (O) indicating the quenching of HPTS-CysMA with 3,3'-oBBV. The fluorimeter settings were as follows: 1% attenuation, ex slit 8 nm, em slit 12 nm, 486 nm ex  $\lambda$ , 537 nm em  $\lambda$ .

HPTS-CysMA ( $1 \times 10^{-5}$  M) and 3,3'-oBBV ( $3 \times 10^{-3}$  M) were titrated with a stock solution of glucose (31250 mg/dL) in pH 7.4 PBS and the fluorescence emission measured after each addition of glucose. The relative change upon addition of glucose is given in FIG. 16.

#### Hydrogel Example

HPTS-CysMA (2 mg), 3,3'-oBBV (15 mg), N,N'-dimethylacrylamide (400 mg), N,N'-methylenebisacrylamide (8 mg), HCl (10  $\mu$ L of 1 M solution), and VA-044 (1 mg) were dissolved in water and diluted to 1 mL in a volumetric flask. The solution was freeze-pump-thawed (3 $\times$ ), injected into a mold containing a 0.005" polyimide spacer and polymerized at 55 $^{\circ}$  C. for 16 h. The resultant film was placed in pH 7.4 phosphate buffer and was tested in a flow cell configuration with increasing amounts of glucose (0, 50, 100, 200, 400 mg/dL). The relative fluorescence change upon addition of glucose is given in FIG. 17. The fluorimeter settings were as follows: ex slit 8 nm, em slit 3.5 nm, 515 nm cutoff filter, 486 nm ex  $\lambda$ , 532 nm em  $\lambda$ .

Although the foregoing invention has been described in terms of certain embodiments and examples, other embodiments will be apparent to those of ordinary skill in the art from the disclosure herein. Moreover, the described embodiments have been presented by way of example only, and are not intended to limit the scope of the inventions. Indeed, the novel methods and systems described herein may be embodied in a variety of other forms without departing from the spirit thereof. Accordingly, other combinations, omissions, substitutions and modifications will be apparent to the skilled artisan in view of the disclosure herein. Thus, the present invention is not intended to be limited by the example or preferred embodiments. The accompanying claims provide exemplary claims and their equivalents are intended to cover forms or modifications as would fall within the scope and spirit of the inventions.

What is claimed is:

1. A sensor for detecting an analyte concentration, comprising:
  - an optical fiber with proximal and distal ends;
  - an atraumatic tip portion with proximal and distal ends, wherein the proximal end of the atraumatic tip portion is separated from the distal end of the optical fiber, such that a gap exists between the atraumatic tip portion and the optical fiber;
  - a rod with proximal and distal ends, wherein the proximal end of the rod is attached to the distal end of the optical fiber, and wherein the distal end of the rod is attached to the proximal end of the atraumatic tip portion, such that the rod traverses the gap and couples the optical fiber to the atraumatic tip portion;
  - a chemical indicator system disposed within the gap and capable of generating an emission light signal in response to an excitation light signal, wherein the intensity of the emission light signal is related to the analyte concentration; and
  - a selectively permeable membrane disposed over the gap.
2. The sensor of claim 1, wherein the chemical indicator system is further immobilized within the gap by a hydrogel.
3. The sensor of claim 1, further comprising a temperature sensor.
4. The sensor of claim 1, wherein the optical fiber has a diameter of between about 0.005 inches and about 0.020 inches.
5. The sensor of claim 1, further comprising a reflective region.
6. The sensor of claim 5, wherein the reflective region comprises a reflective surface of the proximal end of the rod.
7. The sensor of claim 1, wherein the rod is attached to the optical fiber by heating.
8. The sensor of claim 1, wherein the rod is attached to the optical fiber by a reflective adhesive.
9. The sensor of claim 8, wherein the shape of the distal end of the atraumatic tip portion is configured to reduce trauma within tissue and selected from the group consisting of hemispherical, parabolic, and elliptical.
10. The sensor of claim 1, wherein the distal end of the atraumatic tip portion is flexible.
11. The sensor of claim 1, wherein the distal end of the atraumatic tip portion is deformable.
12. The sensor of claim 1, wherein the distal end of the atraumatic tip portion is formed from at least one material selected from the group consisting of plastics, polymers, gels, metals and composites.

13. The sensor of claim 1, wherein the rod is formed from at least one material selected from the group consisting of metal, metal alloy, plastic, polymer, ceramic, and composite material.

14. The sensor of claim 13, wherein the rod is formed from stainless steel, titanium, or Nitinol.

15. The sensor of claim 1, wherein the rod is cylindrical.

16. The sensor of claim 15, wherein the rod diameter is between about 0.002 inches and about 0.010 inches.

17. The sensor of claim 1, wherein the rod is flexible.

18. The sensor of claim 1, wherein the rod is stiffer than the optical fiber.

19. The sensor of claim 1, wherein the rod is sufficiently stiff to prevent flexing of the sensor along the gap.

20. The sensor of claim 1, wherein the sensor is sized for deployment within tissue.

21. A sensor for detecting an analyte concentration, comprising:

an optical fiber with proximal and distal ends;

an atraumatic tip portion with proximal and distal ends, wherein the proximal end of the atraumatic tip portion is separated from the distal end of the optical fiber, such that a gap exists between the atraumatic tip portion and the optical fiber;

a hypotube with proximal and distal ends, wherein the proximal end of the hypotube is attached to the distal end of the optical fiber, and wherein the distal end of the hypotube is attached to the proximal end of the atraumatic tip portion, such that the hypotube traverses the gap and couples the optical fiber to the atraumatic tip portion, wherein the hypotube comprises at least one window that opens onto the gap;

a chemical indicator system disposed within the gap and capable of generating an emission light signal in response to an excitation light signal, wherein the intensity of the emission light signal is related to the analyte concentration; and

a selectively permeable membrane disposed over the at least one window.

22. The sensor of claim 21, wherein the chemical indicator system is further immobilized within the gap by a hydrogel.

23. The sensor of claim 21, wherein the sensor is sized for deployment within tissue.

24. A sensor for detecting an analyte concentration, comprising:

an optical fiber with proximal and distal ends;

an atraumatic tip portion with proximal and distal ends, wherein the proximal end of the atraumatic tip portion is separated from the distal end of the optical fiber, such that a gap exists between the atraumatic tip portion and the optical fiber;

a cage connecting the optical fiber and atraumatic tip portion, wherein the optical fiber is at least partially enclosed within the cage, and wherein the cage has at least one window;

a chemical indicator system disposed within the cage, wherein the chemical indicator system is adjacent the window and is separated from analyte by a selectively permeable membrane, and wherein the chemical indicator system is capable of generating an emission light signal in response to an excitation light signal, wherein the intensity of the emission light signal is related to the analyte concentration; and

a reference material, wherein the reference material is configured to either reflect a portion of the excitation light signal before the excitation light signal enters the chemical indicator system or to return a second emission light signal, wherein the intensity of the second emission light signal is not related to the analyte concentration.

25. The sensor of claim 24, wherein the sensor is sized for deployment within tissue.

26. A sensor for detecting an analyte concentration, comprising:

an optical fiber with a proximal end and a distal end, wherein the distal end of the optical fiber comprises a glucose sensing hydrogel and wherein the glucose sensing hydrogel comprises a first fluorophore, a quencher, and at least one glucose binding moiety;

a reference fiber adjacent the optical fiber having a proximal end and a distal end, wherein the distal end of the reference fiber comprises a reference material, wherein the reference material comprises a second fluorophore;

a light emitting diode operably coupled to the glucose fiber and the reference fiber, wherein the light emitting diode sends an excitation light to the glucose fiber and the reference fiber;

a glucose signal detector operatively coupled to the glucose fiber, wherein the glucose signal detector receives a first fluorescent light from the glucose fiber; and

a reference signal detector operatively coupled to the reference fiber, wherein the reference signal detector receives a second fluorescent light from the reference fiber.

27. The sensor of claim 26, wherein the sensor is sized for deployment within tissue.

28. The sensor of claim 26, wherein the first fluorophore is the same as the second fluorophore.

29. The sensor of claim 26, wherein the first fluorophore is different than the second fluorophore.

30. The sensor of claim 26, wherein the reference material comprises a quencher.

31. The sensor of claim 26, wherein the reference material is encased in a glucose impermeable membrane.

专利名称(译)	用于比率校正葡萄糖测量的光学传感器配置		
公开(公告)号	<a href="#">US8700115</a>	公开(公告)日	2014-04-15
申请号	US13/894718	申请日	2013-05-15
[标]申请(专利权)人(译)	GLUMETRICS		
申请(专利权)人(译)	GLUMETRICS INC.		
当前申请(专利权)人(译)	MEDTRONIC MINIMED INC.		
[标]发明人	MARKLE DAVID R GAMSEY SOYA PEYSER THOMAS A		
发明人	MARKLE, DAVID R. GAMSEY, SOYA PEYSER, THOMAS A.		
IPC分类号	A61B5/00		
CPC分类号	A61B5/14735 A61B2562/043 A61B5/1459 G01N2021/7786 A61B5/14532 A61B2562/0238 G01N21/7703		
其他公开文献	US20130267802A1		
外部链接	<a href="#">Espacenet</a> <a href="#">USPTO</a>		

摘要(译)

本发明的实施例涉及一种用于检测葡萄糖的光学传感器。传感器包括设置在光纤远端和无创伤尖端部分之间的间隙内的化学指示器系统，其中光纤和无创伤尖端部分通过耦合构件耦合，例如穿过的杆或海波管或笼子。差距。传感器还包括用于产生和检测与葡萄糖无关的光学参考信号的装置，使得能够对光学系统中的伪像进行比率校正葡萄糖测量。

