US010258265B1

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

(10) **Patent No.: US 10,258,265 B1**
(45) **Date of Patent: *Apr. 16, 2019**

(54) **MULTI-STREAM DATA COLLECTION SYSTEM FOR NONINVASIVE MEASUREMENT OF BLOOD CONSTITUENTS**

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(*) 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.: **16/212,440**

(22) Filed: **Dec. 6, 2018**

Related U.S. Application Data

(63) Continuation of application No. 14/981,290, filed on Dec. 28, 2015, which is a continuation of application (Continued)

(51) **Int. Cl.**

A61B 5/1455	(2006.01)
A61B 5/00	(2006.01)
A61B 5/145	(2006.01)

(52) **U.S. Cl.**
CPC ..... ***A61B 5/1455*** (2013.01); ***A61B 5/14532*** (2013.01); ***A61B 5/14546*** (2013.01); (Continued)

(58) **Field of Classification Search**
CPC . A61B 5/0205; A61B 5/1455; A61B 5/14551; A61B 5/14552; A61B 5/14532; (Continued)

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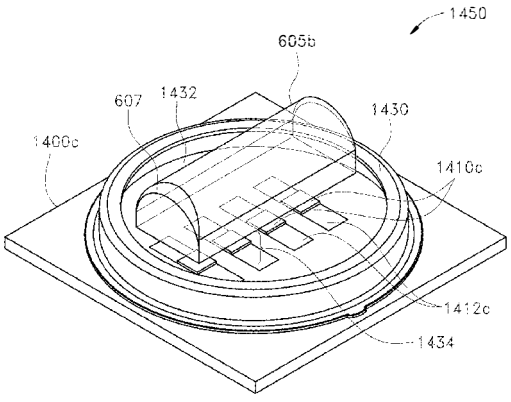
(57) **ABSTRACT**

The present disclosure relates to noninvasive methods, devices, and systems for measuring various blood constituents or analytes, such as glucose. In an embodiment, a light source comprises LEDs and super-luminescent LEDs. The light source emits light at least wavelengths of about 1610 nm, about 1640 nm, and about 1665 nm. In an embodiment, the detector comprises a plurality of photodetectors arranged in a special geometry comprising one of a substantially (Continued)



linear substantially equal spaced geometry, a substantially linear substantially non-equal spaced geometry, and a substantially grid geometry.

30 Claims, 65 Drawing Sheets

Related U.S. Application Data

No. 12/829,352, filed on Jul. 1, 2010, now Pat. No. 9,277,880, which is a continuation of application No. 12/534,827, filed on Aug. 3, 2009, now abandoned, and a continuation-in-part of application No. 12/497,528, filed on Jul. 2, 2009, now Pat. No. 8,577,431, which is a continuation-in-part of application No. 29/323,409, filed on Aug. 25, 2008, now Pat. No. Des. 621,516, and a continuation-in-part of application No. 29/323,408, filed on Aug. 25, 2008, now Pat. No. Des. 606,659, said application No. 12/829,352 is a continuation-in-part of application No. 12/497,523, filed on Jul. 2, 2009, now Pat. No. 8,437,825, which is a continuation-in-part of application No. 29/323,409, and a continuation-in-part of application No. 29/323,408.

(60) Provisional application No. 61/086,060, filed on Aug. 4, 2008, provisional application No. 61/086,108, filed on Aug. 4, 2008, provisional application No. 61/086,063, filed on Aug. 4, 2008, provisional application No. 61/086,057, filed on Aug. 4, 2008, provisional application No. 61/091,732, filed on Aug. 25, 2008, provisional application No. 61/078,228, filed on Jul. 3, 2008, provisional application No. 61/078,207, filed on Jul. 3, 2008.

(52) **U.S. Cl.**
CPC ***A61B 5/14552*** (2013.01); ***A61B 5/6816*** (2013.01); ***A61B 5/6826*** (2013.01); ***A61B 5/6829*** (2013.01); ***A61B 5/6838*** (2013.01); ***A61B 5/6843*** (2013.01); *A61B 2562/0233* (2013.01); *A61B 2562/04* (2013.01); *A61B 2562/046* (2013.01); *A61B 2562/146* (2013.01)

(58) **Field of Classification Search**
CPC . A61B 5/14546; A61B 5/6829; A61B 5/6843; A61B 5/6826; A61B 5/6816; A61B 5/6838
See application file for complete search history.

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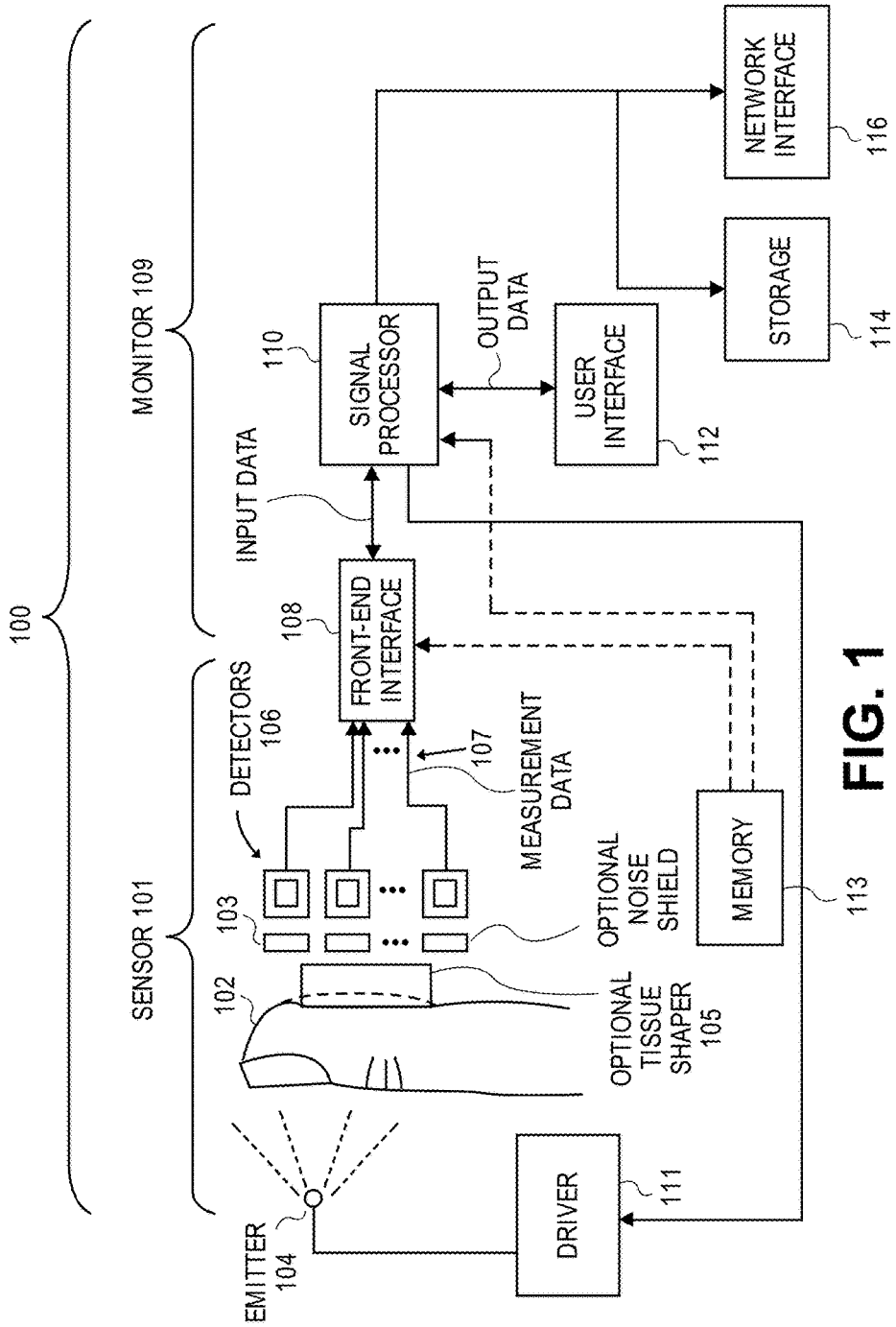

100
MONITOR 109
SENSOR 101
NETWORK INTERFACE
116
STORAGE
114
SIGNAL PROCESSOR
110
OUTPUT DATA
USER INTERFACE
112
INPUT DATA
FRONT-END INTERFACE
108
DETECTORS
106
107
MEASUREMENT DATA
OPTIONAL NOISE SHIELD
MEMORY
113
103
102
OPTIONAL TISSUE SHAPER
105
DRIVER
111
EMITTER
104

FIG. 1

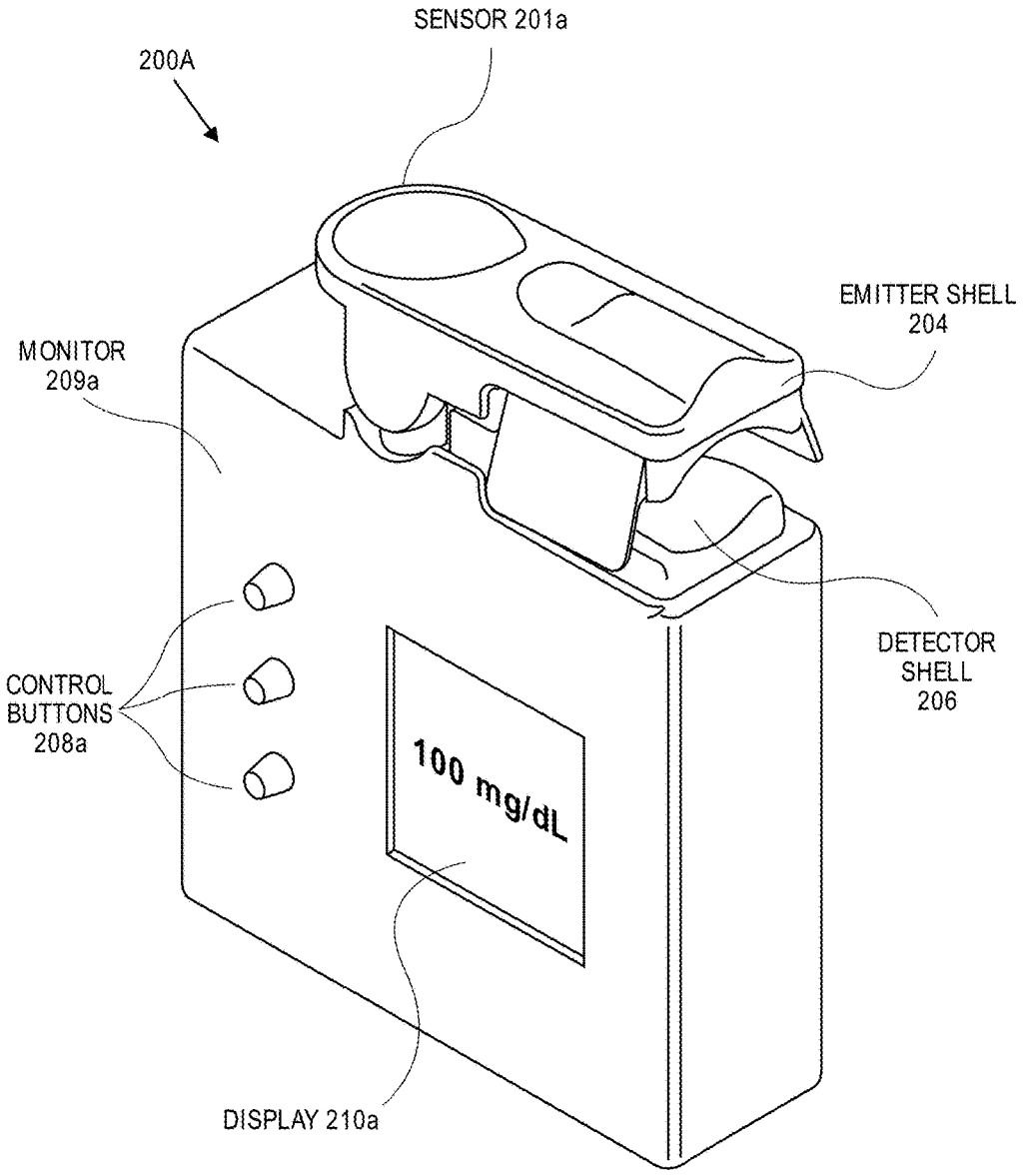

SENSOR 201a
200A
EMITTER SHELL
204
MONITOR
209a
DETECTOR
SHELL
206
CONTROL
BUTTONS
208a
100 mg/dL
DISPLAY 210a

FIG. 2A

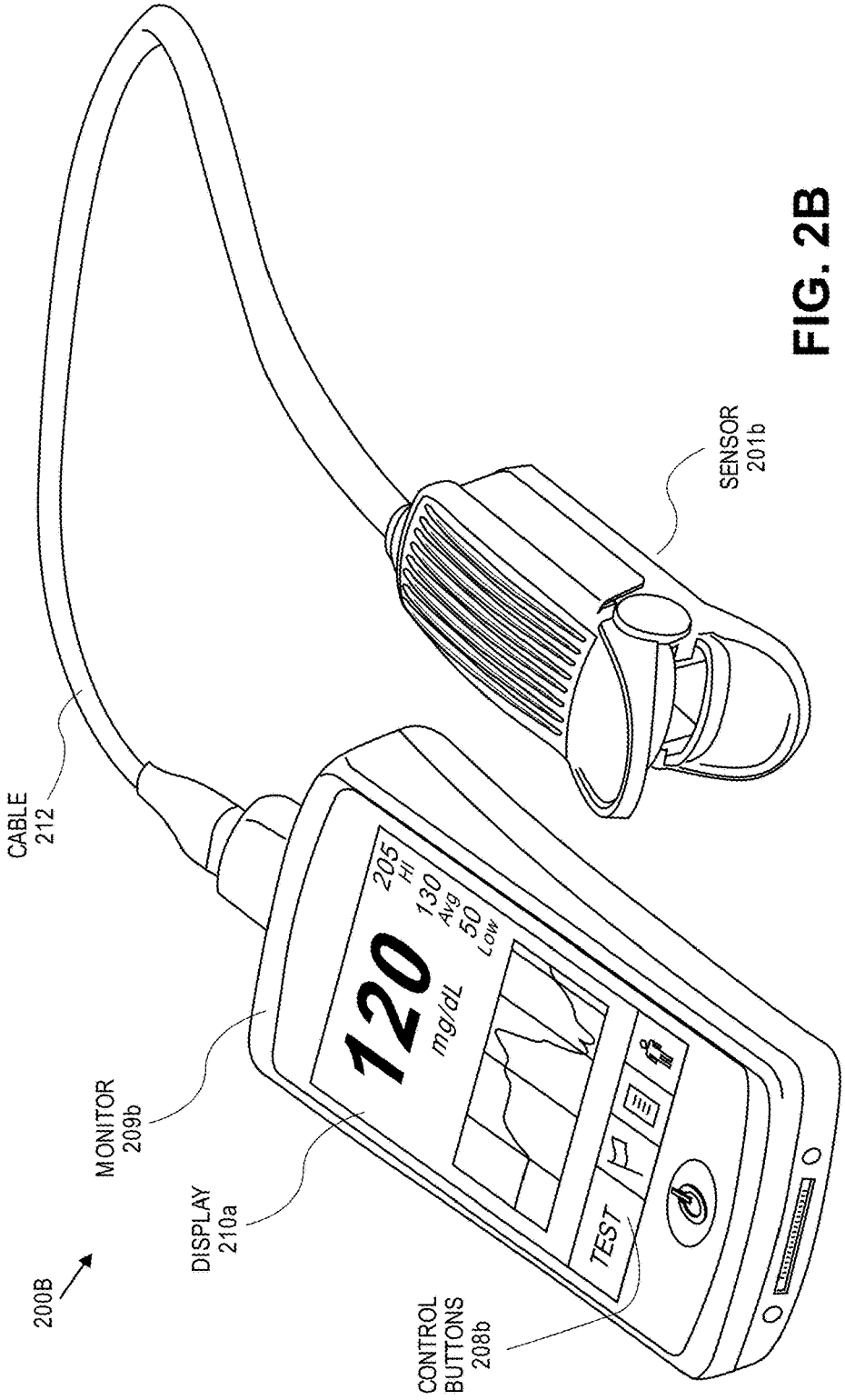

CABLE
212
SENSOR
201b
MONITOR
209b
DISPLAY
210a
200B
CONTROL
BUTTONS
208b
205
Hi
130
Avg
50
Low
120
mg/dL
TEST

FIG. 2B

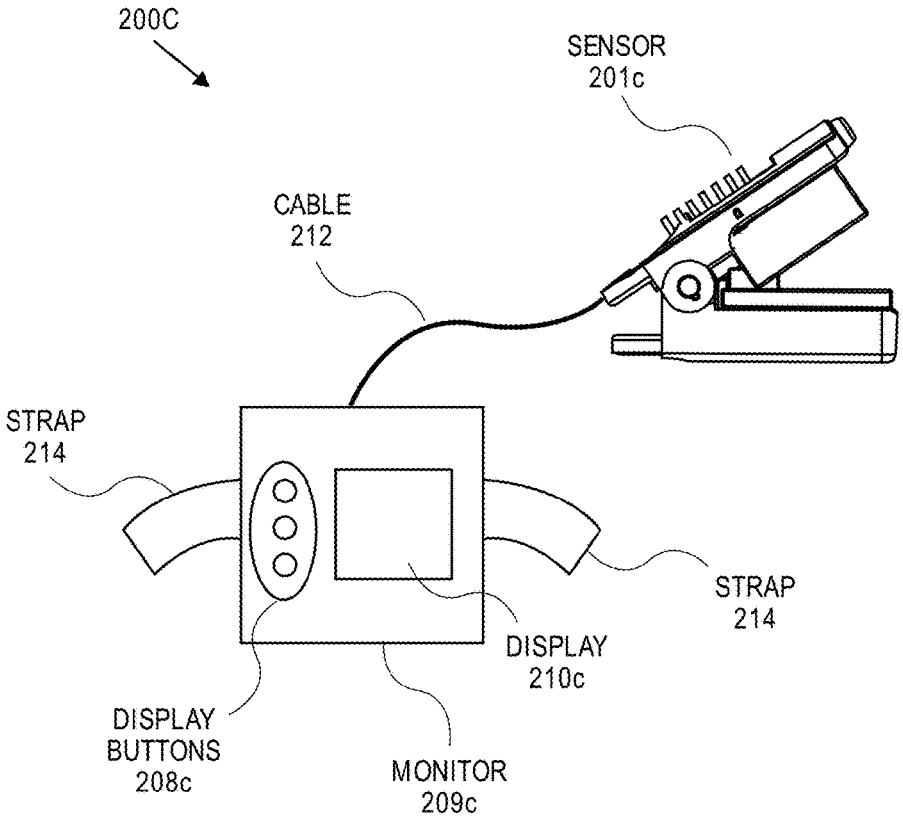

200C
SENSOR
201c
CABLE
212
STRAP
214
STRAP
214
DISPLAY
210c
DISPLAY
BUTTONS
208c
MONITOR
209c

FIG. 2C

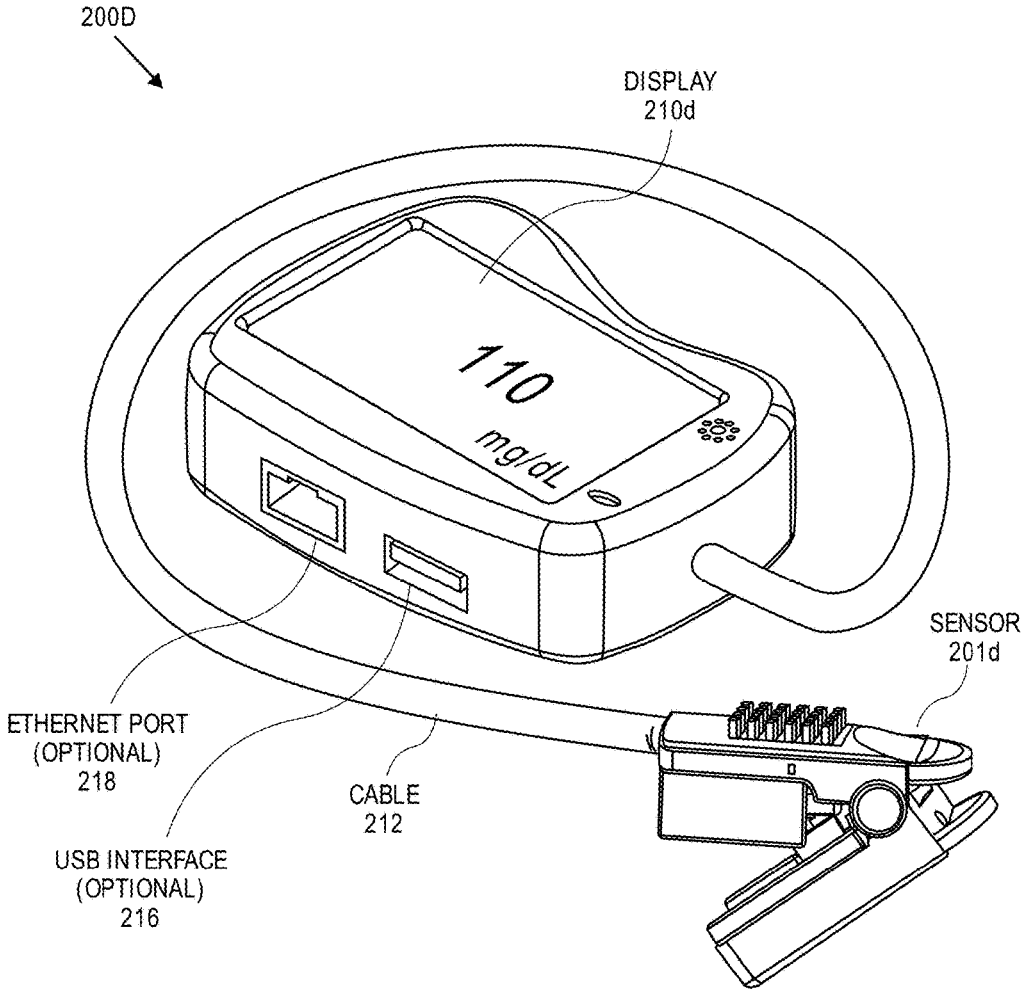

200D
DISPLAY
210d
110
mg/dL
SENSOR
201d
ETHERNET PORT
(OPTIONAL)
218
CABLE
212
USB INTERFACE
(OPTIONAL)
216

FIG. 2D

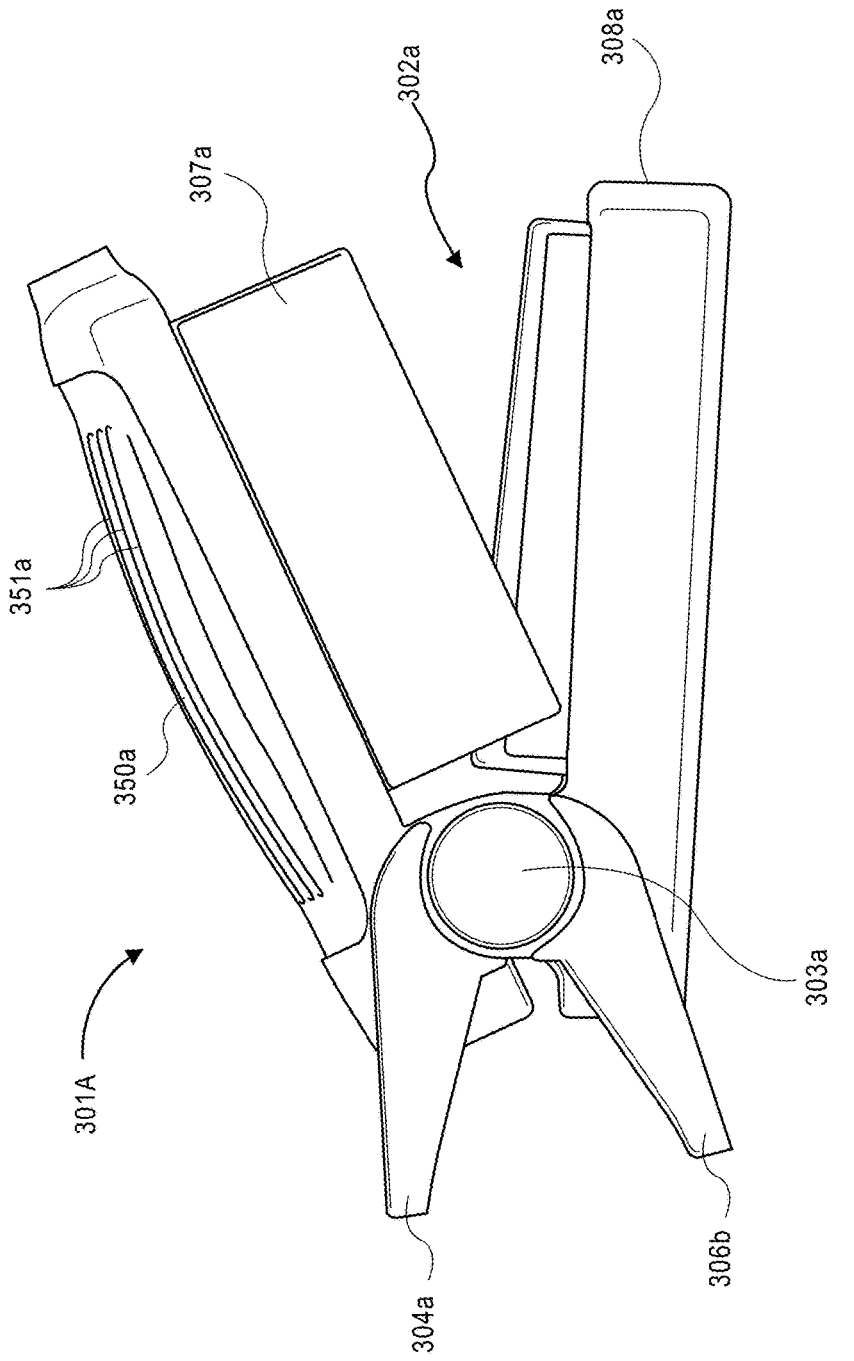


FIG. 3A

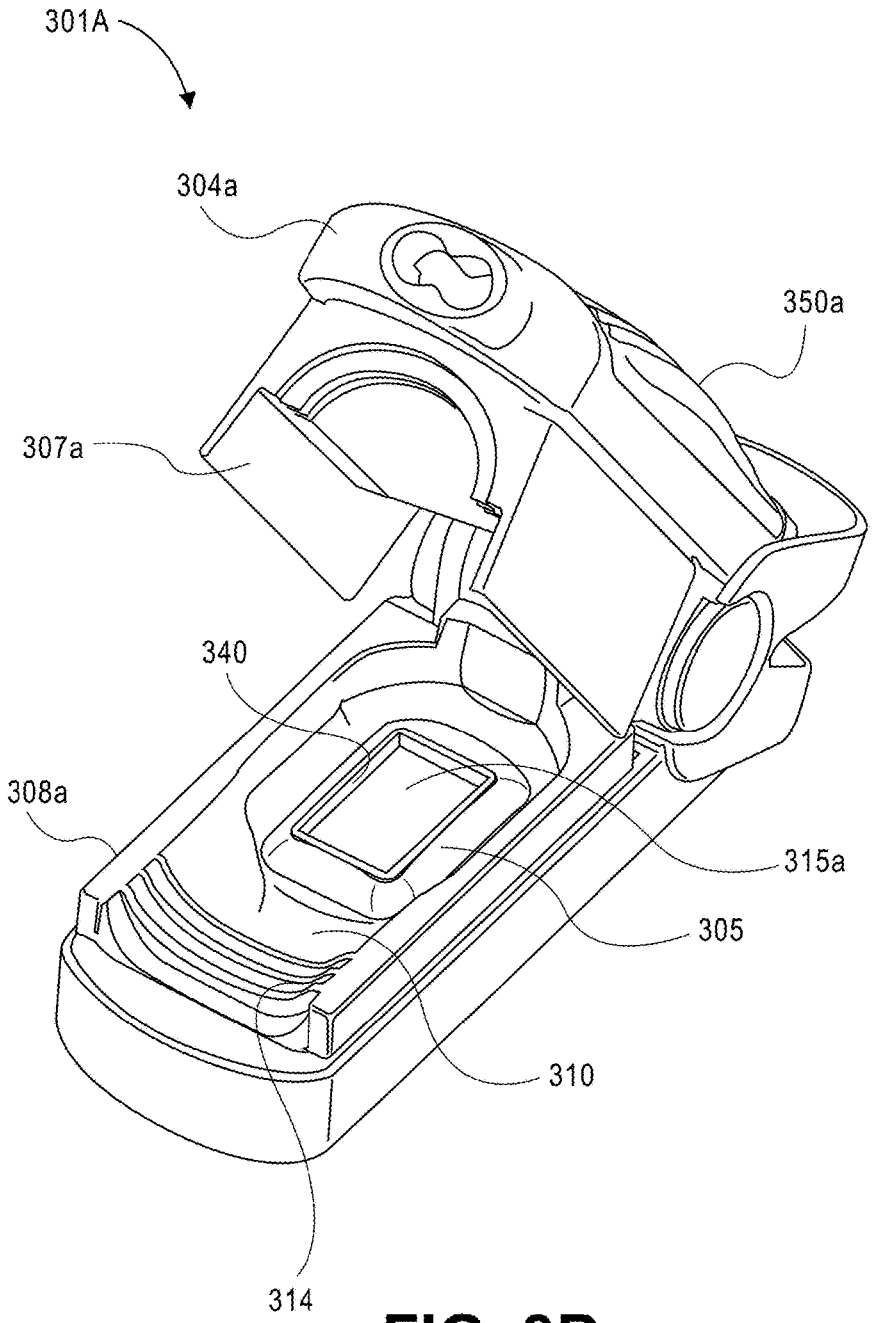


FIG. 3B

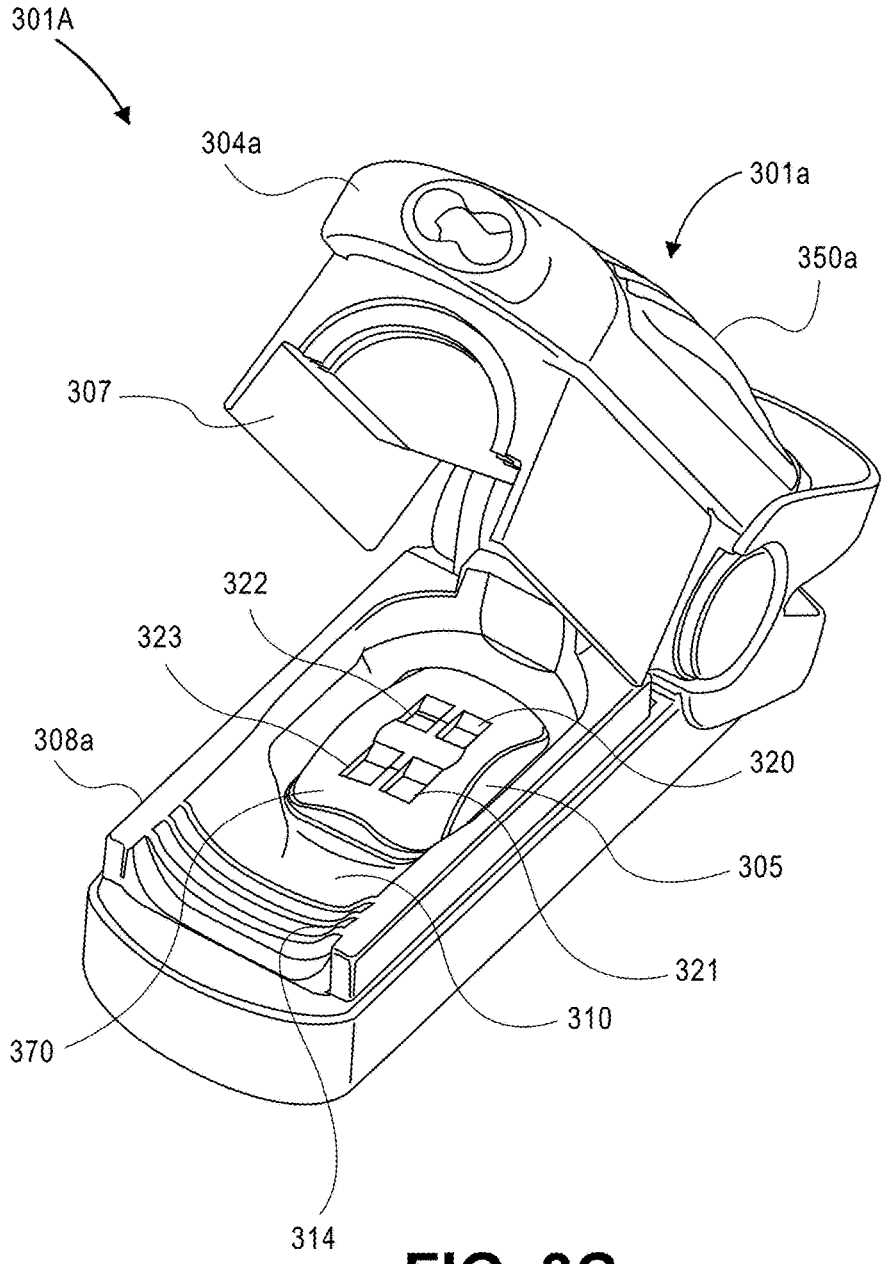


FIG. 3C

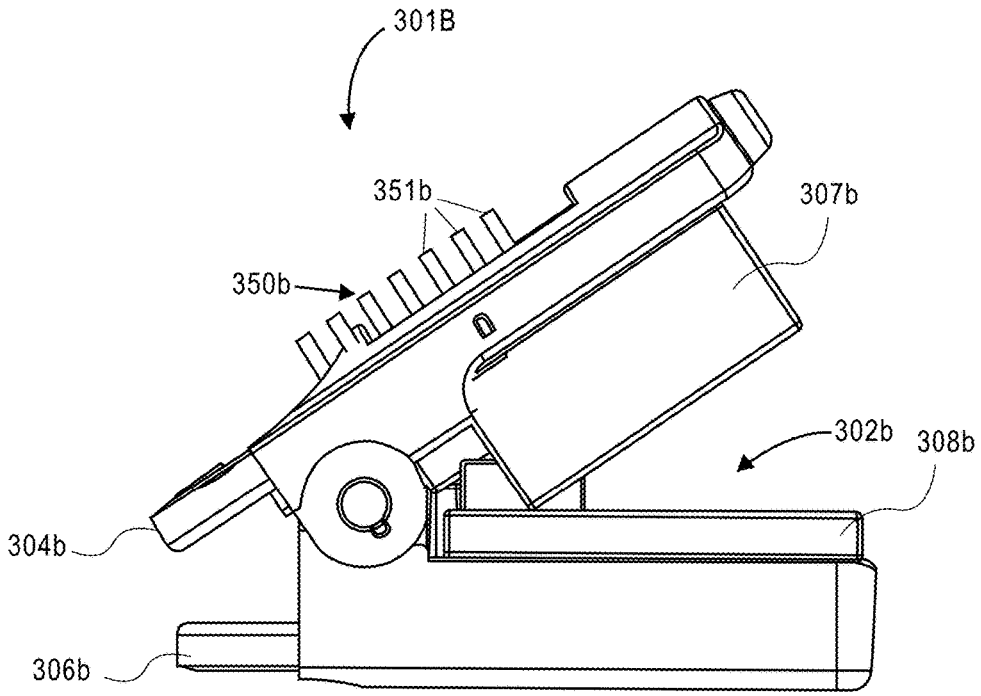


FIG. 3D

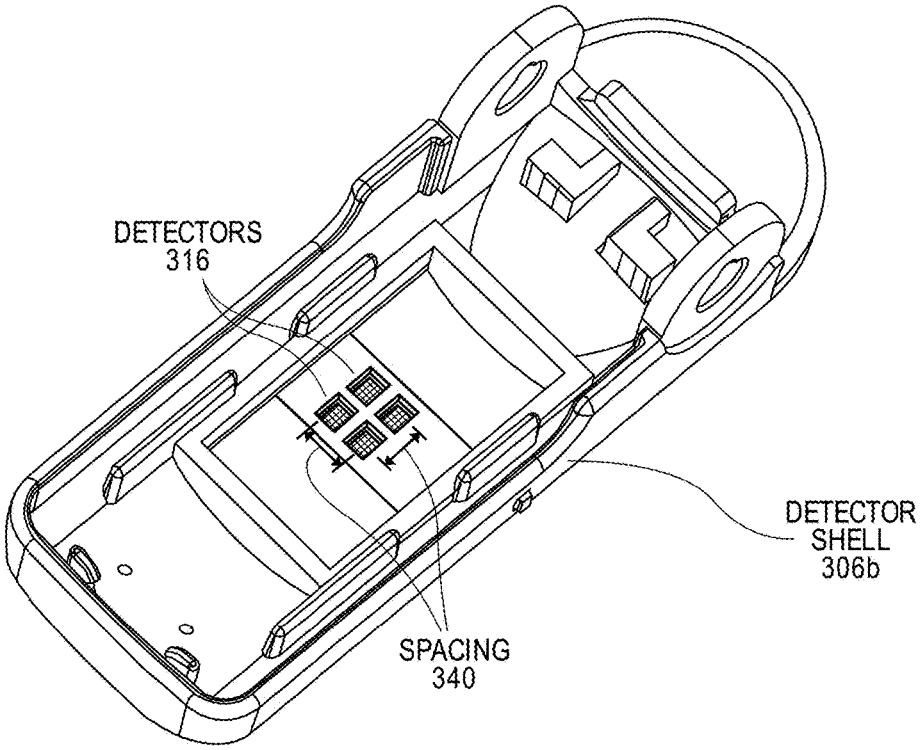


FIG. 3E

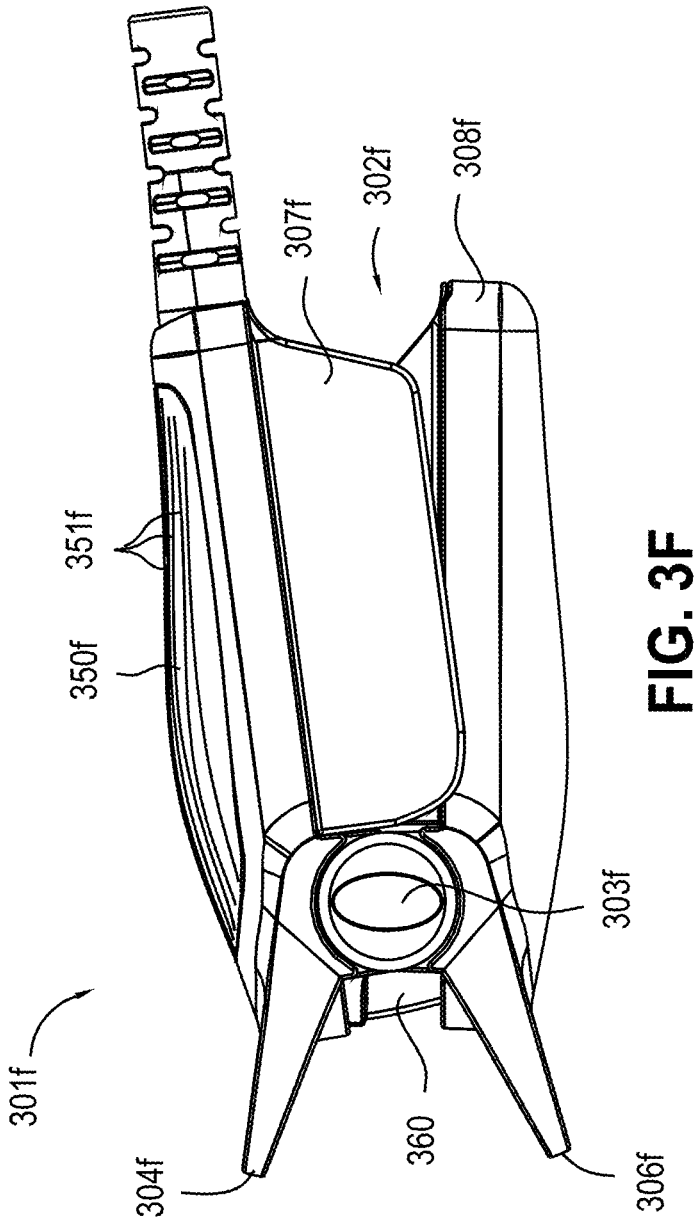


FIG. 3F

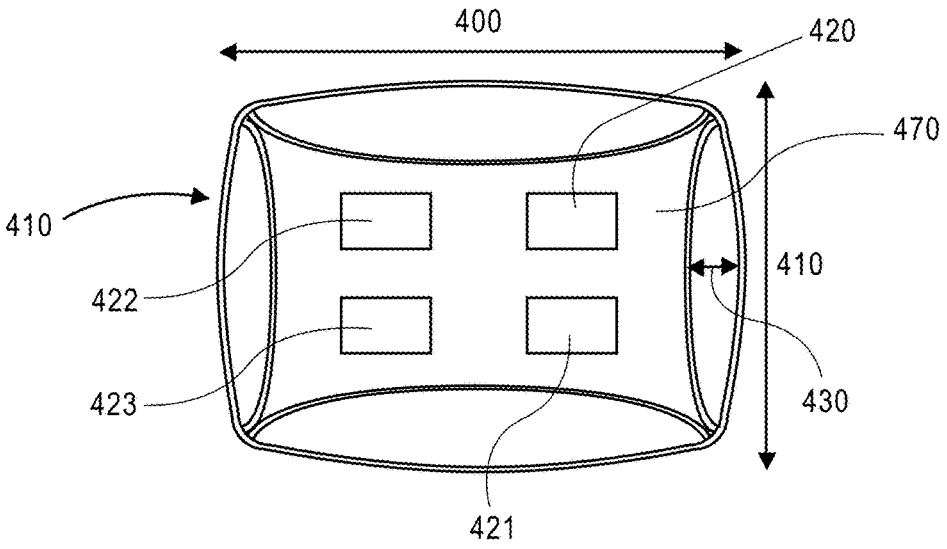


FIG. 4A

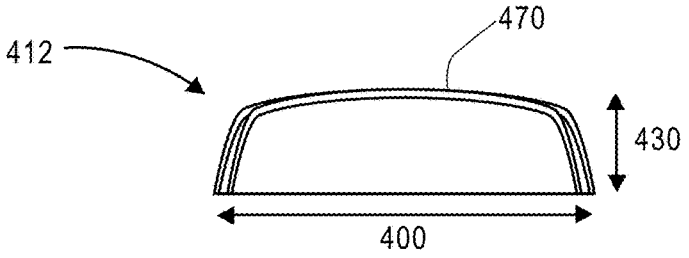


FIG. 4B

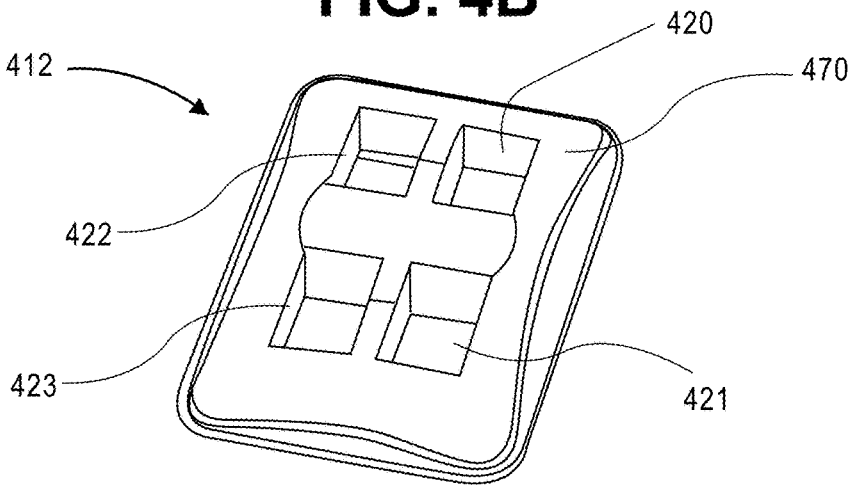


FIG. 4C

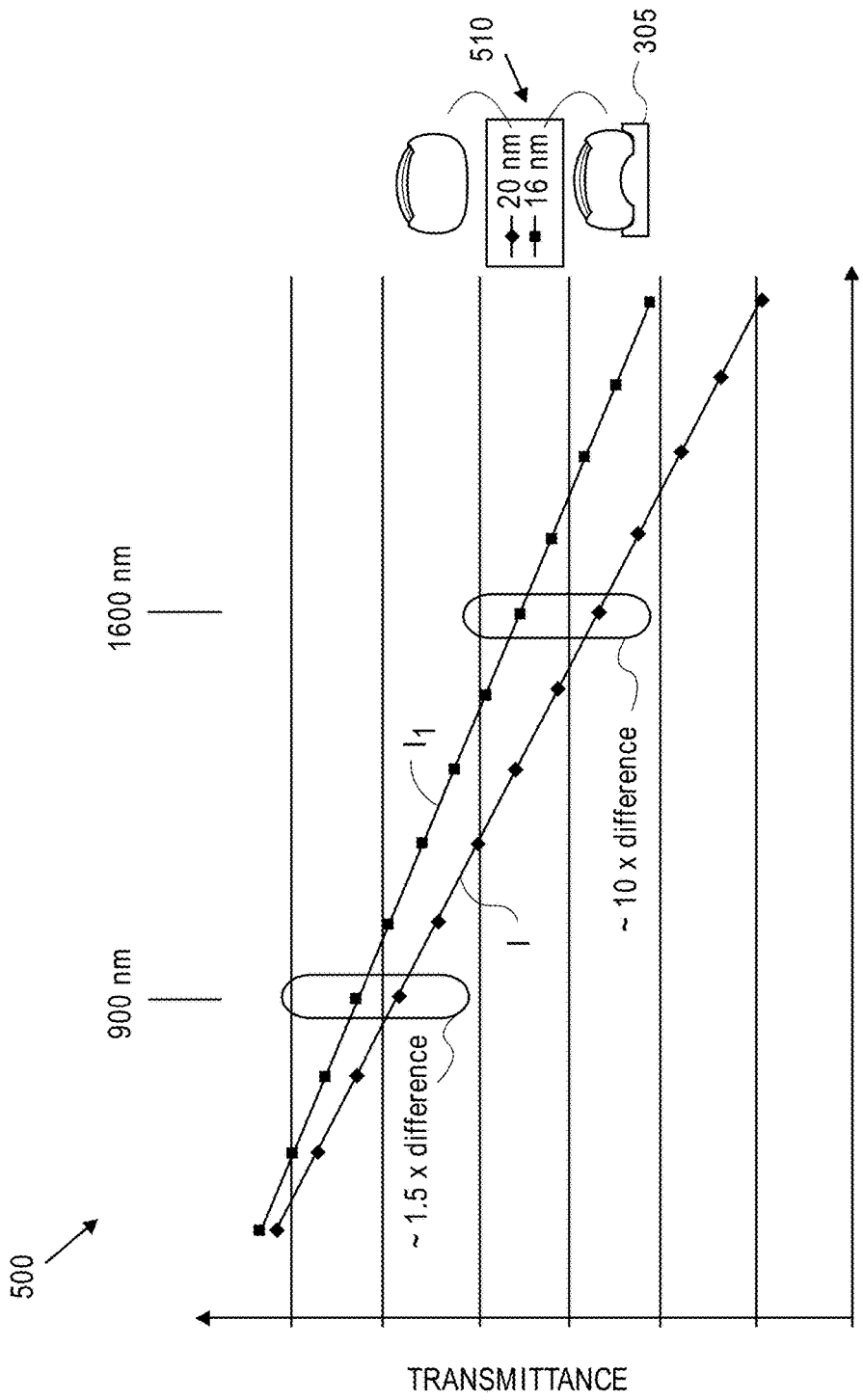

510
305
20 nm
16 nm
1600 nm
900 nm
I_1
I
~ 1.5 x difference
~ 10 x difference
500
TRANSMITTANCE
WAVELENGTH

FIG. 5

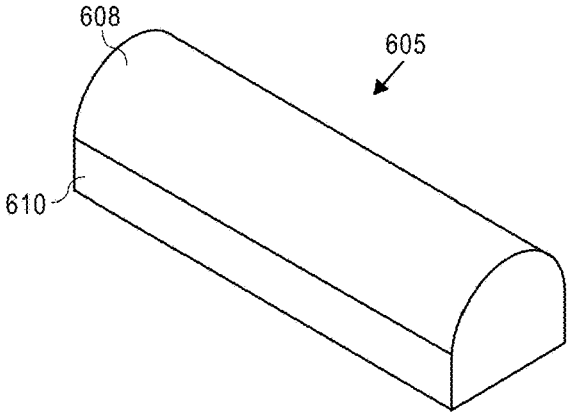


FIG. 6A

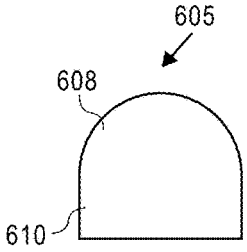


FIG. 6B

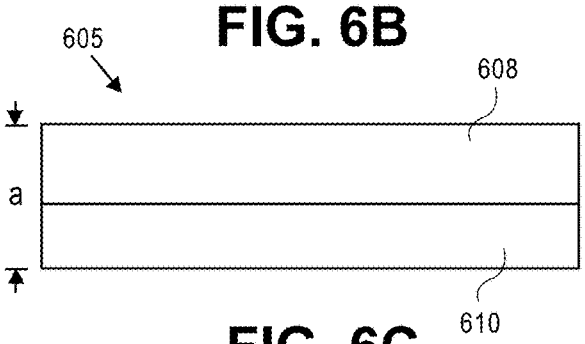


FIG. 6C

FIG. 6D

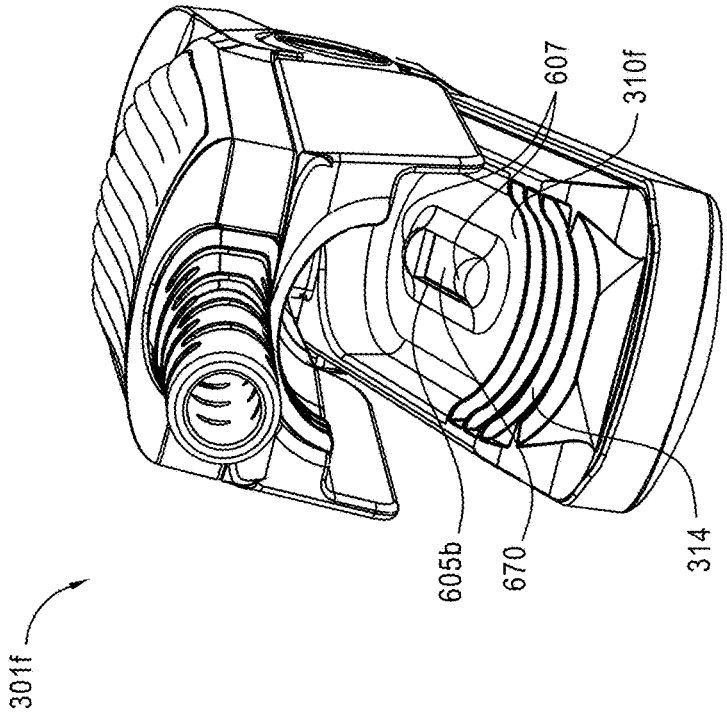


FIG. 6E

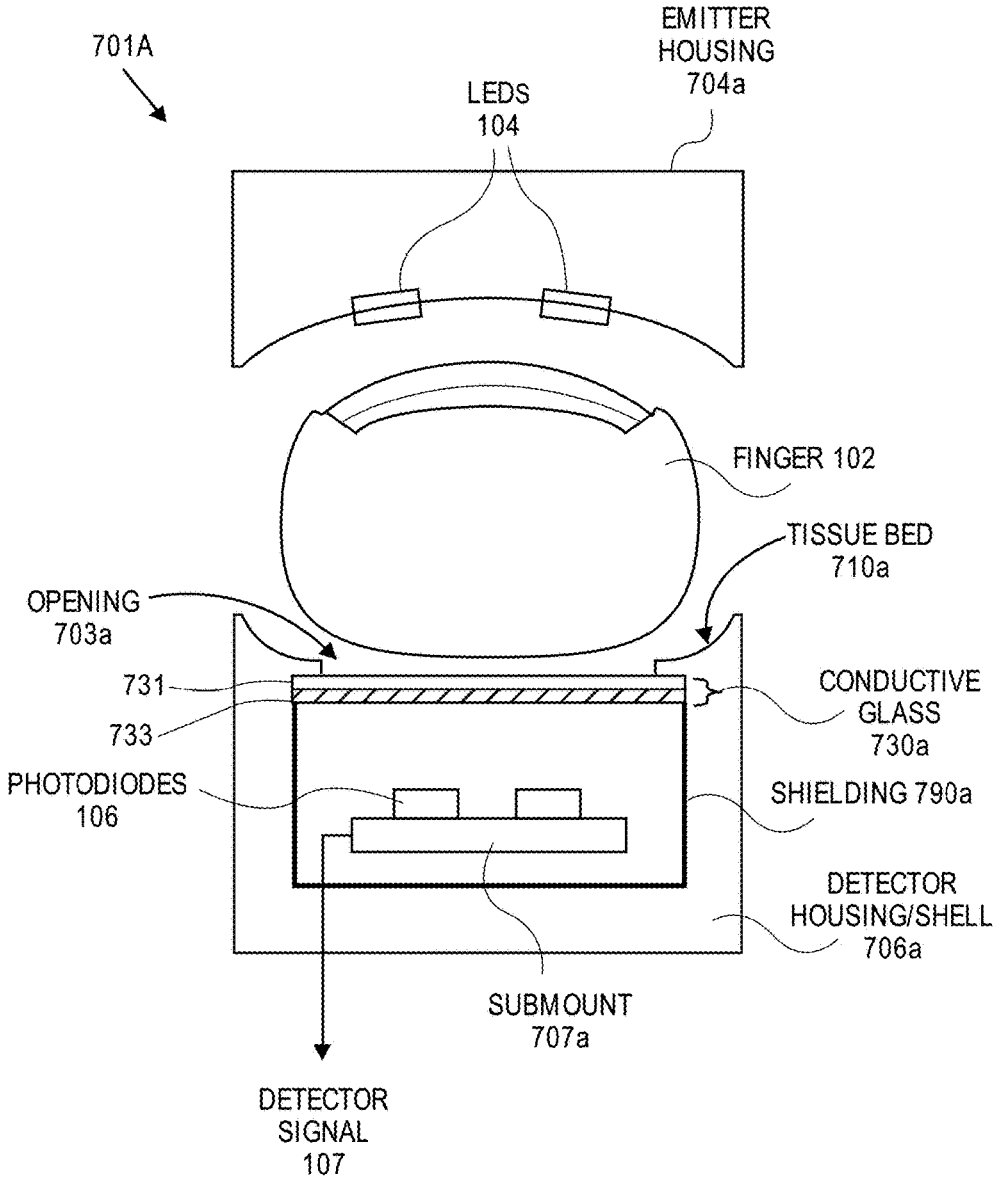

701A
EMITTER HOUSING 704a
LEDS 104
FINGER 102
TISSUE BED 710a
OPENING 703a
731
733
CONDUCTIVE GLASS 730a
PHOTODIODES 106
SHIELDING 790a
DETECTOR HOUSING/SHELL 706a
SUBMOUNT 707a
DETECTOR SIGNAL 107

FIG. 7A

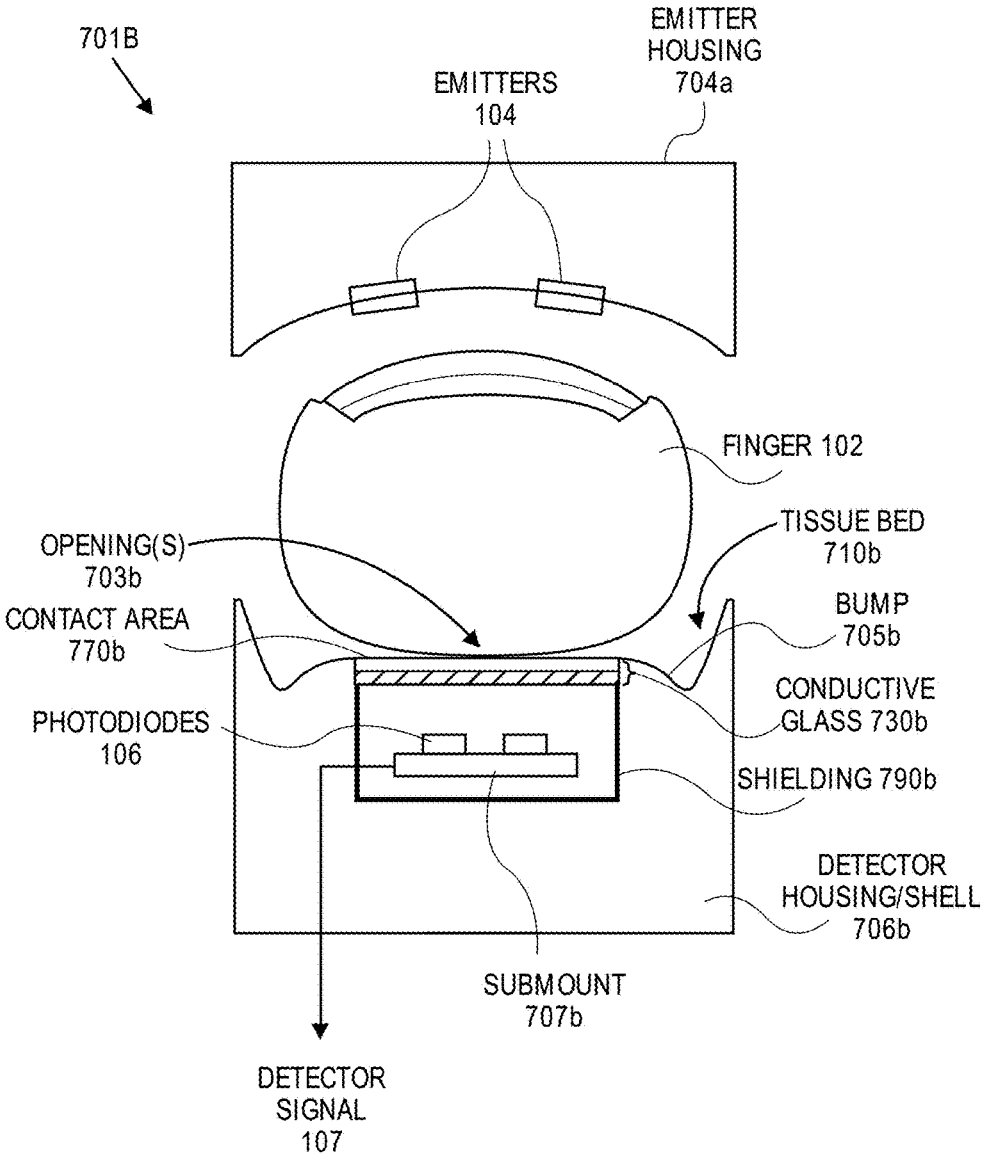

701B
EMITTER HOUSING 704a
EMITTERS 104
FINGER 102
TISSUE BED 710b
OPENING(S) 703b
CONTACT AREA 770b
BUMP 705b
CONDUCTIVE GLASS 730b
PHOTODIODES 106
SHIELDING 790b
DETECTOR HOUSING/SHELL 706b
SUBMOUNT 707b
DETECTOR SIGNAL 107

FIG. 7B

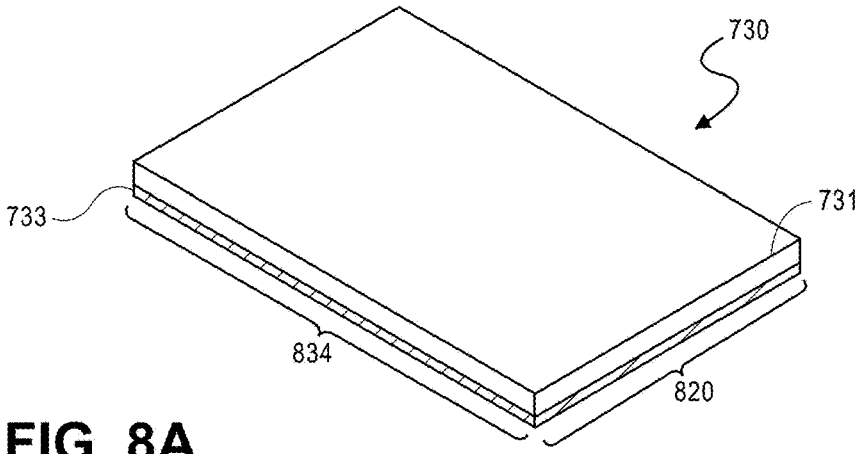


FIG. 8A

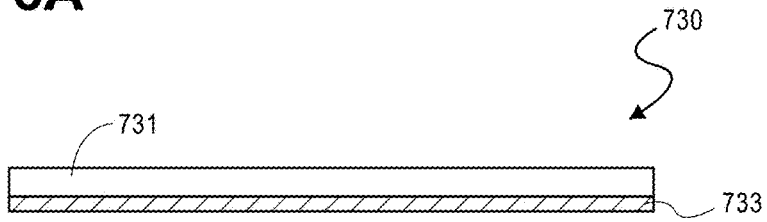


FIG. 8B

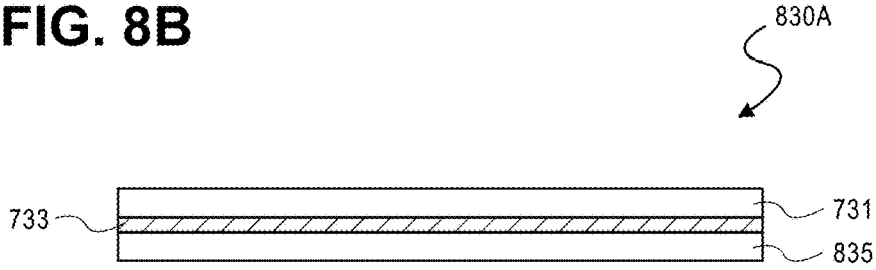


FIG. 8C

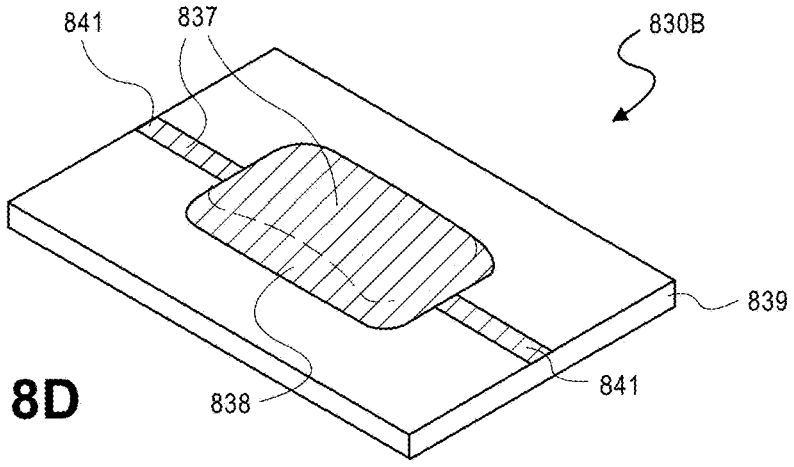


FIG. 8D

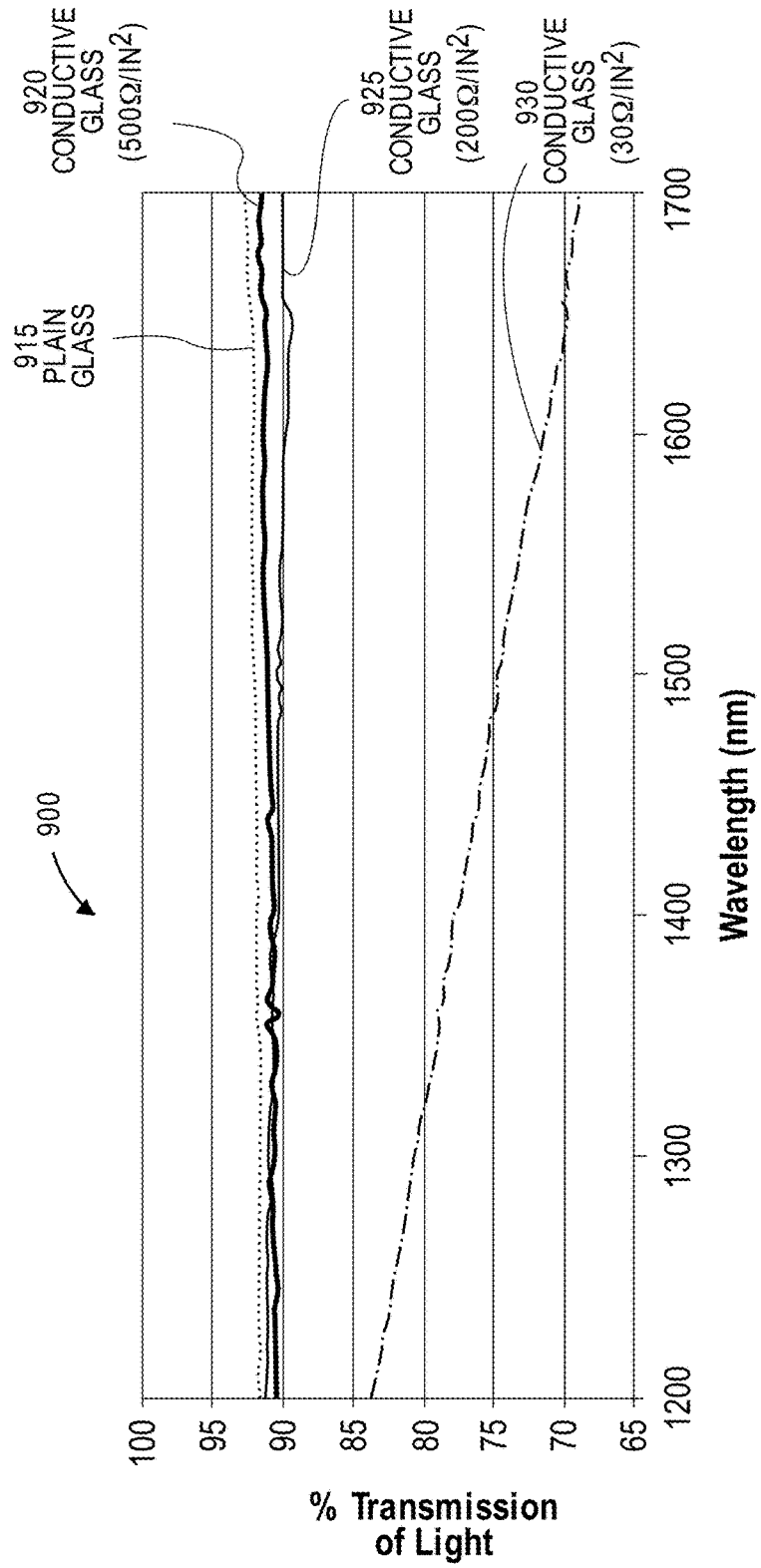

920 CONDUCTIVE GLASS (500Ω/IN²)
925 CONDUCTIVE GLASS (200Ω/IN²)
930 CONDUCTIVE GLASS (30Ω/IN²)
915 PLAIN GLASS
900
100
95
90
85
80
75
70
65
% Transmission of Light
1200
1300
1400
1500
1600
1700
Wavelength (nm)

FIG. 9

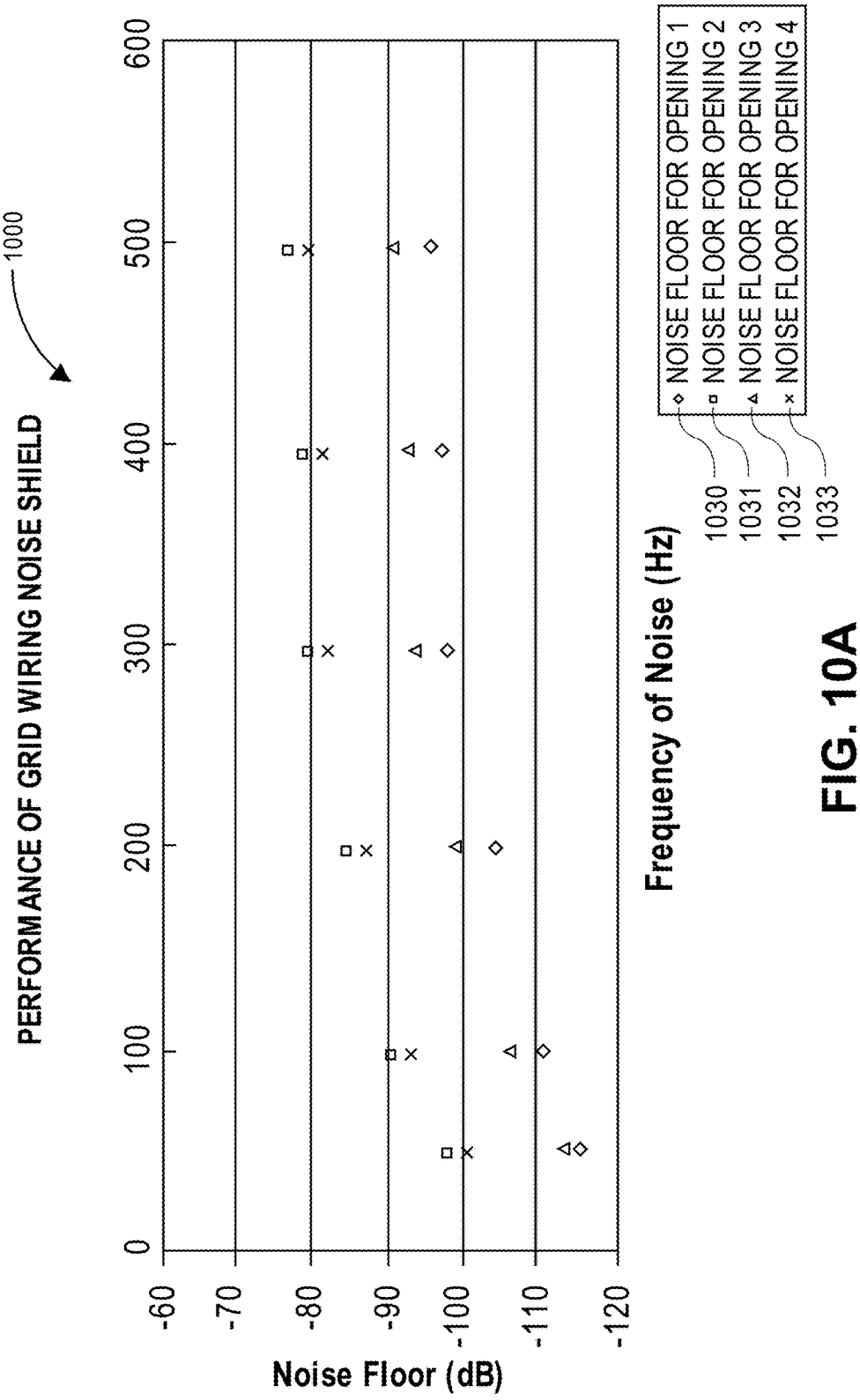

1000
PERFORMANCE OF GRID WIRING NOISE SHIELD
Noise Floor (dB)
-60
-70
-80
-90
-100
-110
-120
Frequency of Noise (Hz)
0
100
200
300
400
500
600
NOISE FLOOR FOR OPENING 1
NOISE FLOOR FOR OPENING 2
NOISE FLOOR FOR OPENING 3
NOISE FLOOR FOR OPENING 4
1030
1031
1032
1033

FIG. 10A

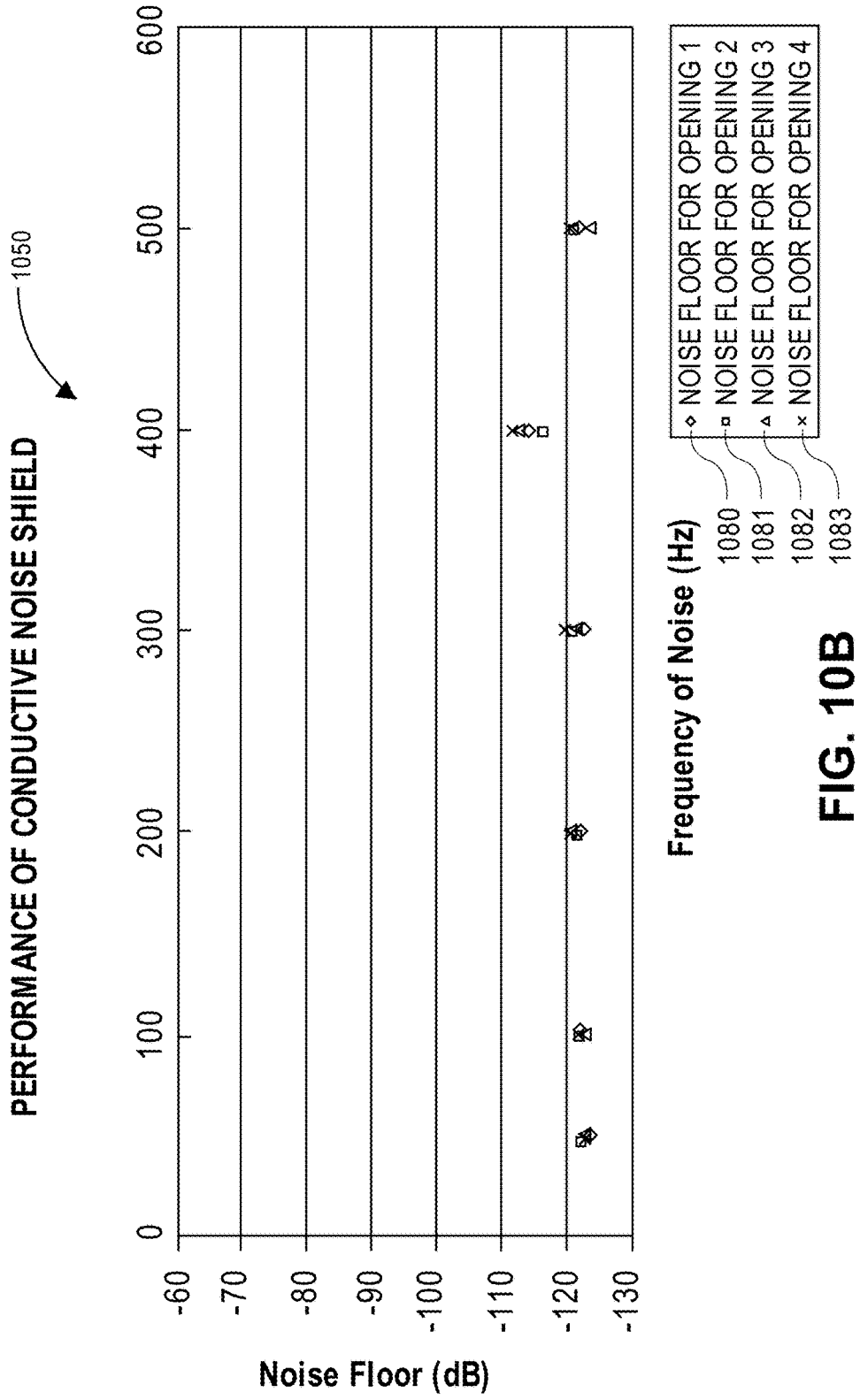

PERFORMANCE OF CONDUCTIVE NOISE SHIELD
1050
Noise Floor (dB)
-60
-70
-80
-90
-100
-110
-120
-130
Frequency of Noise (Hz)
0
100
200
300
400
500
600
NOISE FLOOR FOR OPENING 1
NOISE FLOOR FOR OPENING 2
NOISE FLOOR FOR OPENING 3
NOISE FLOOR FOR OPENING 4
1080
1081
1082
1083

FIG. 10B

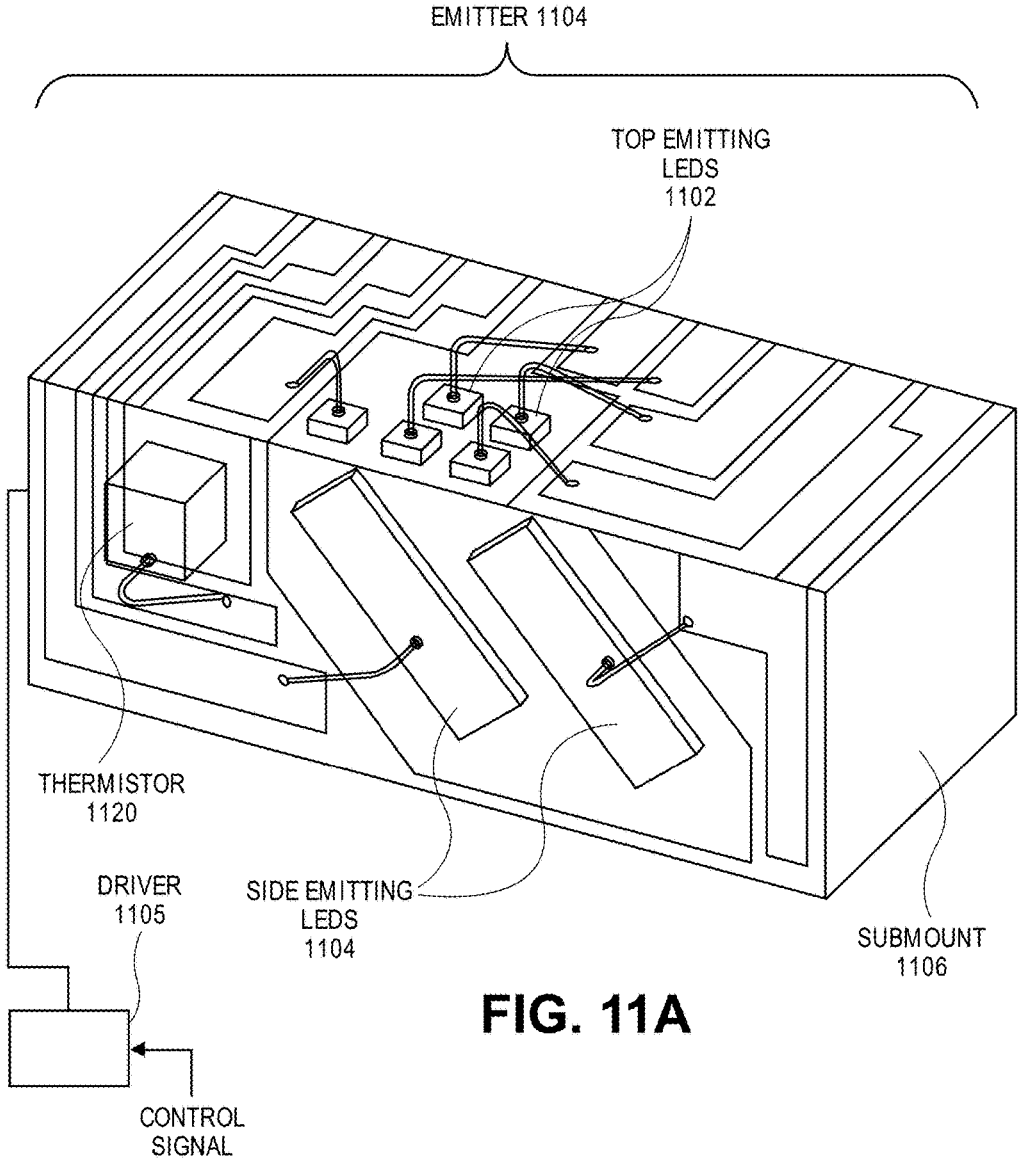

EMITTER 1104
TOP EMITTING LEDS 1102
THERMISTOR 1120
DRIVER 1105
SIDE EMITTING LEDS 1104
SUBMOUNT 1106
CONTROL SIGNAL

FIG. 11A

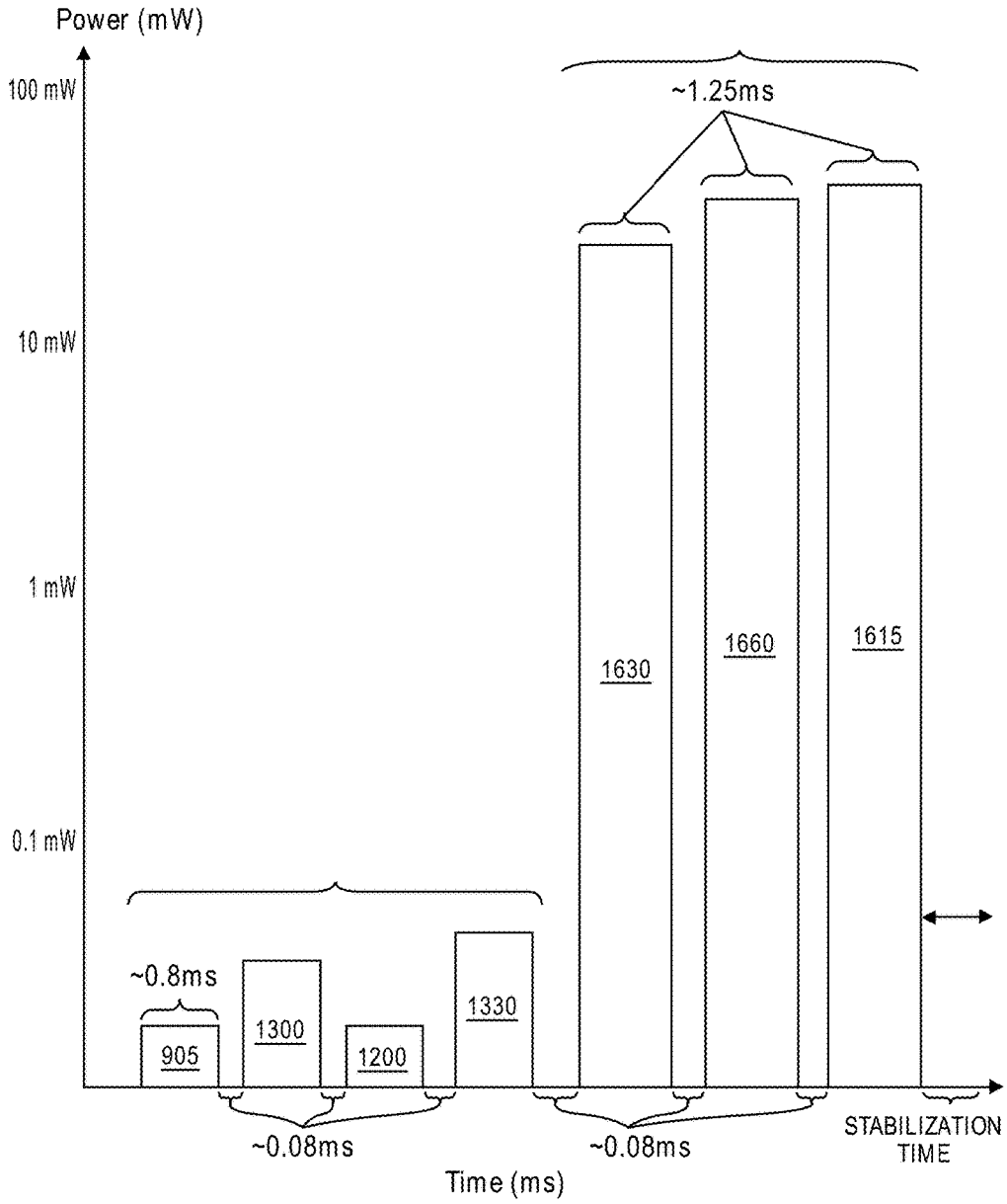

Power (mW)
100 mW
10 mW
1 mW
0.1 mW
~1.25ms
1630
1660
1615
~0.8ms
905
1300
1200
1330
~0.08ms
~0.08ms
STABILIZATION TIME
Time (ms)

FIG. 11B

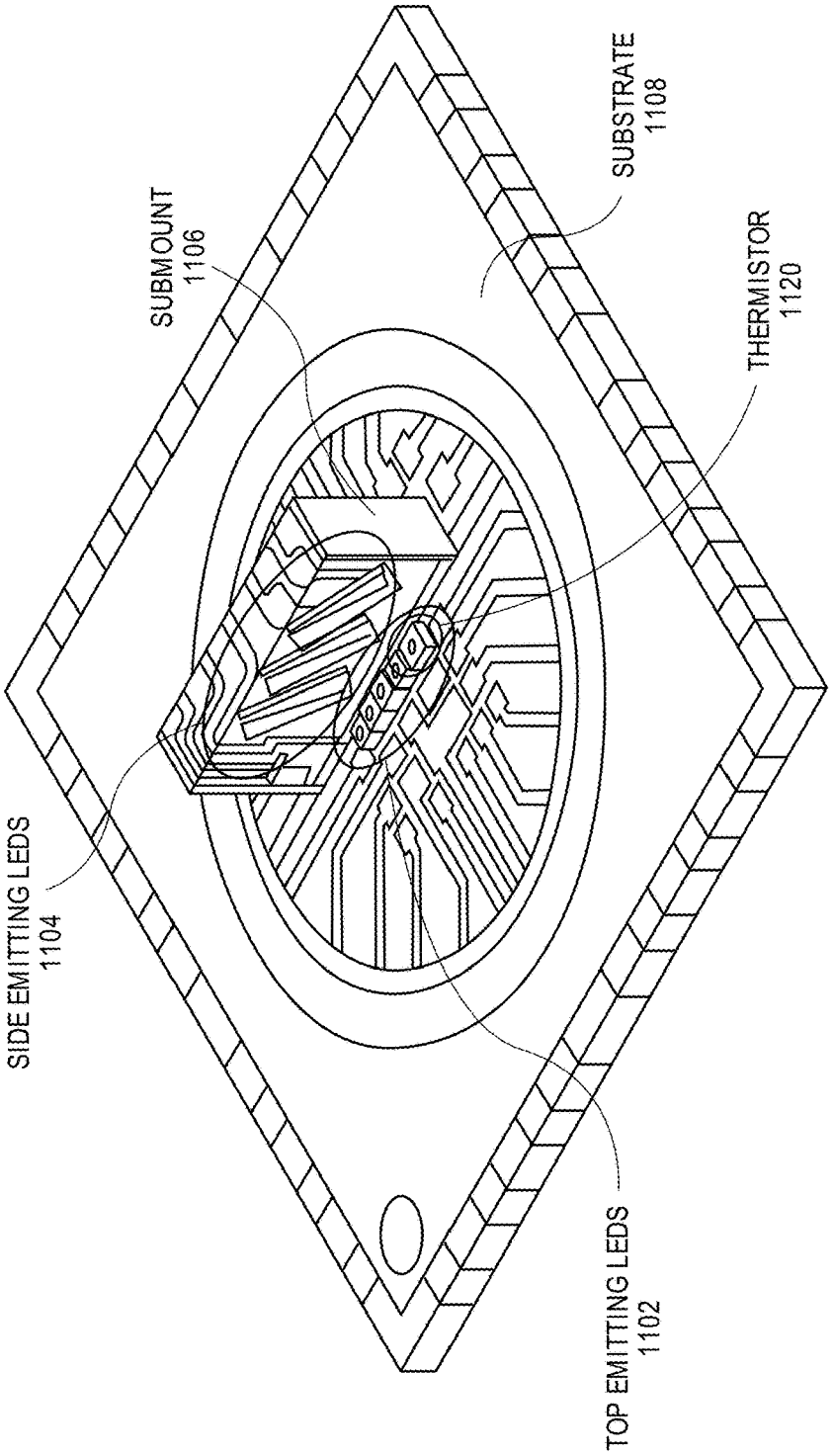

SUBSTRATE
1108
SUBMOUNT
1106
THERMISTOR
1120
SIDE EMITTING LEDS
1104
TOP EMITTING LEDS
1102

FIG. 11C

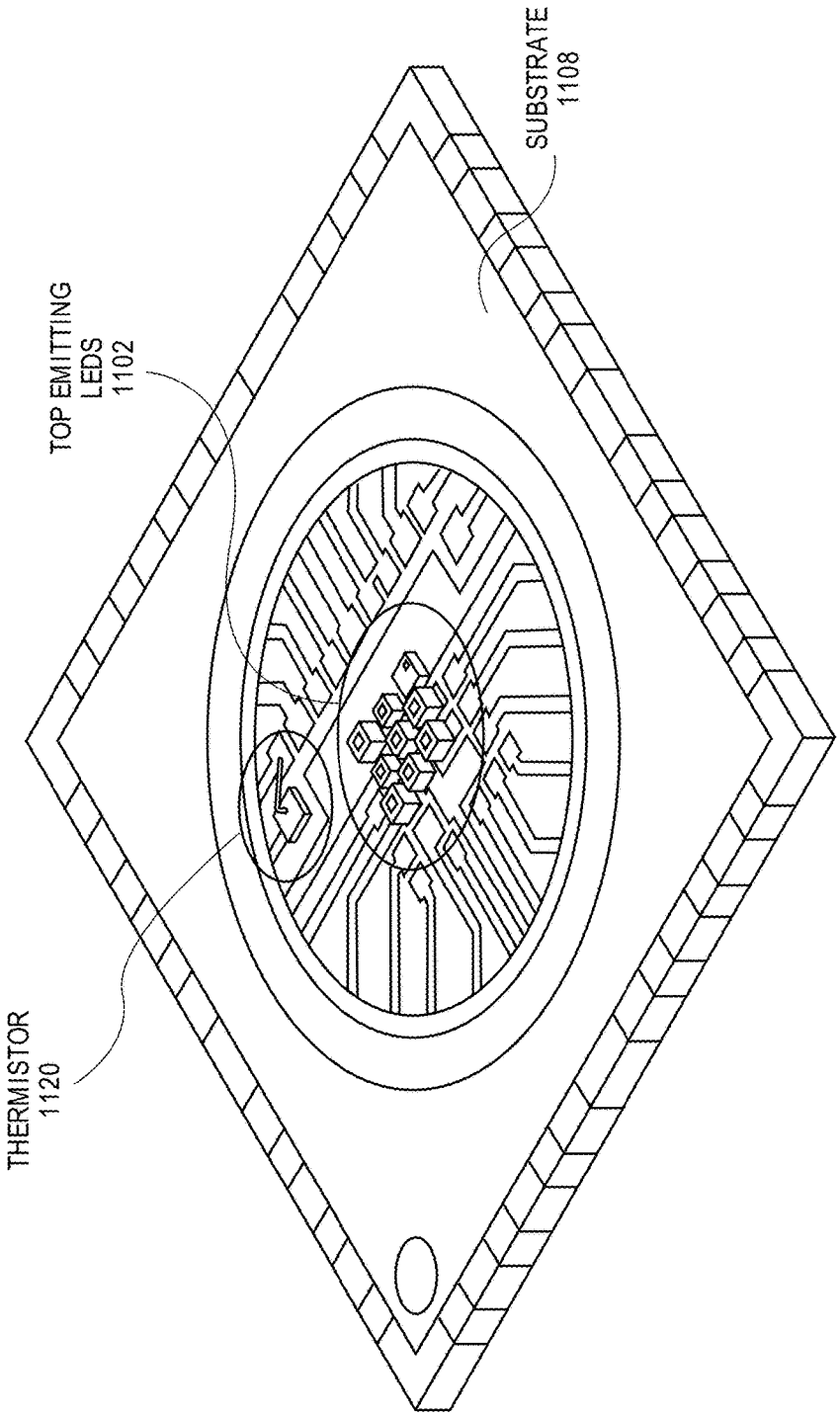


FIG. 11D

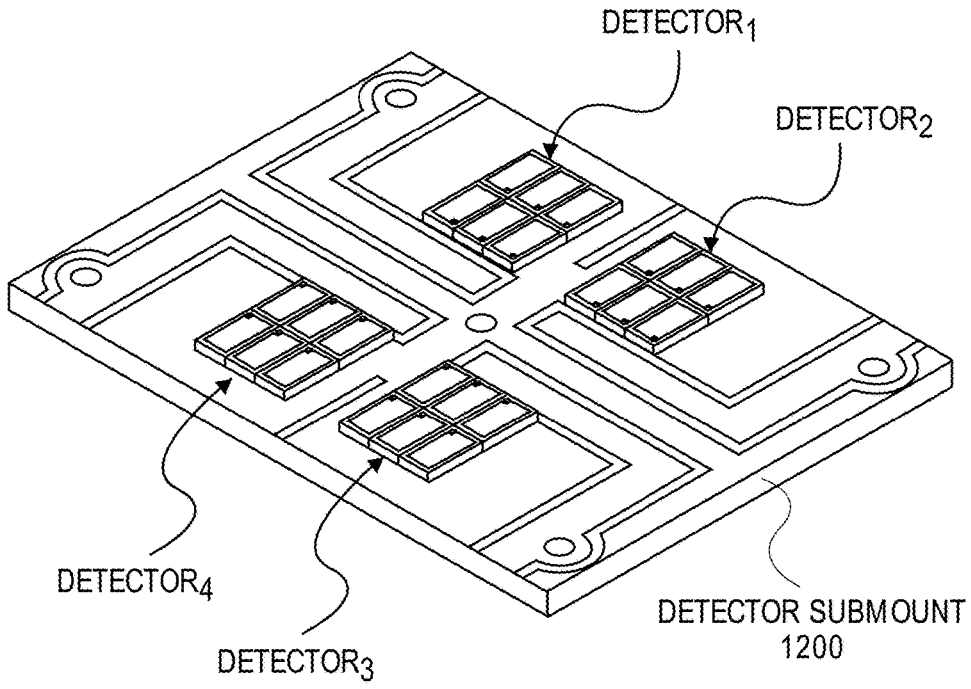

DETECTOR1
DETECTOR2
DETECTOR4
DETECTOR3
DETECTOR SUBMOUNT
1200

FIG. 12A

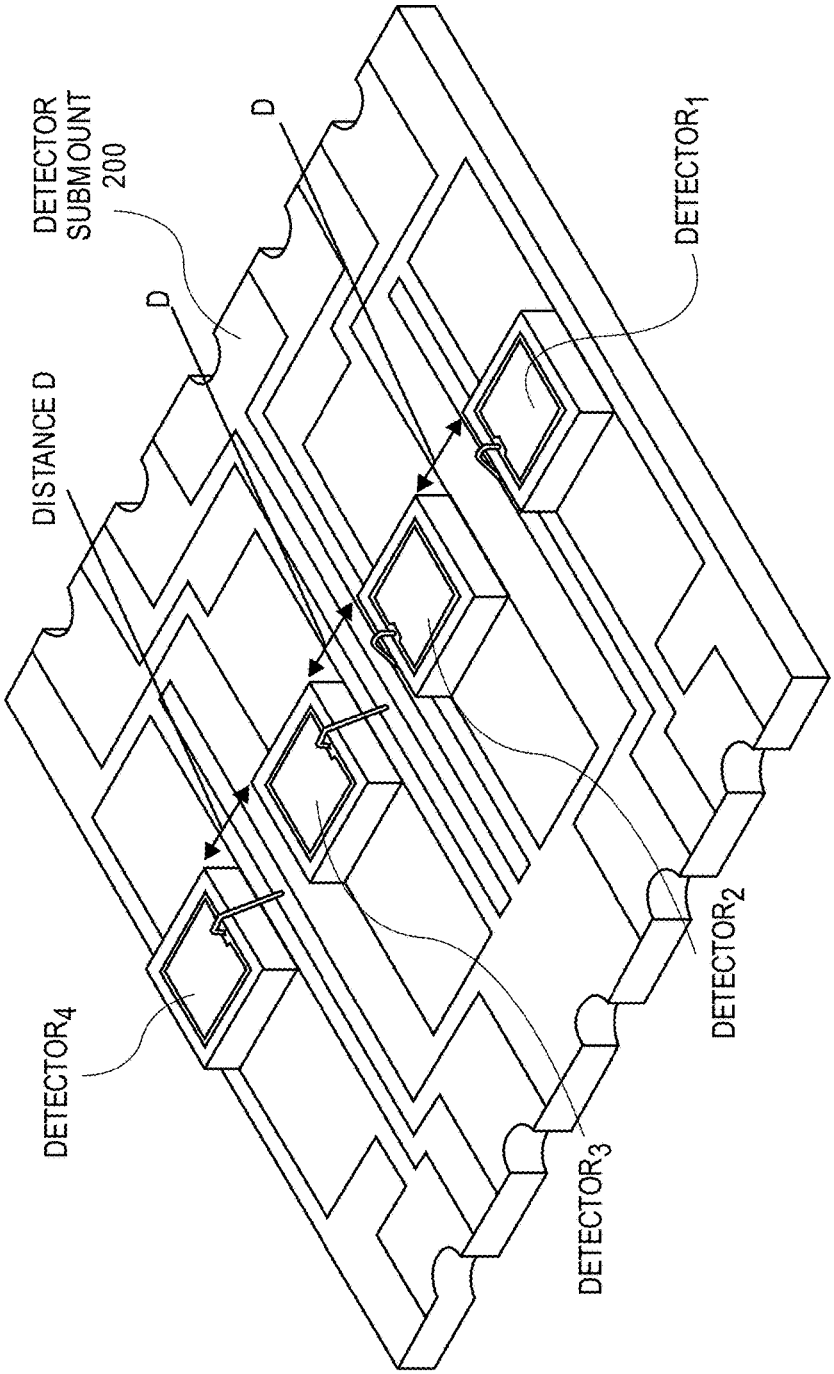

DETECTOR SUBMOUNT 200
D
D
DISTANCE D
DETECTOR1
DETECTOR2
DETECTOR3
DETECTOR4

FIG. 12B

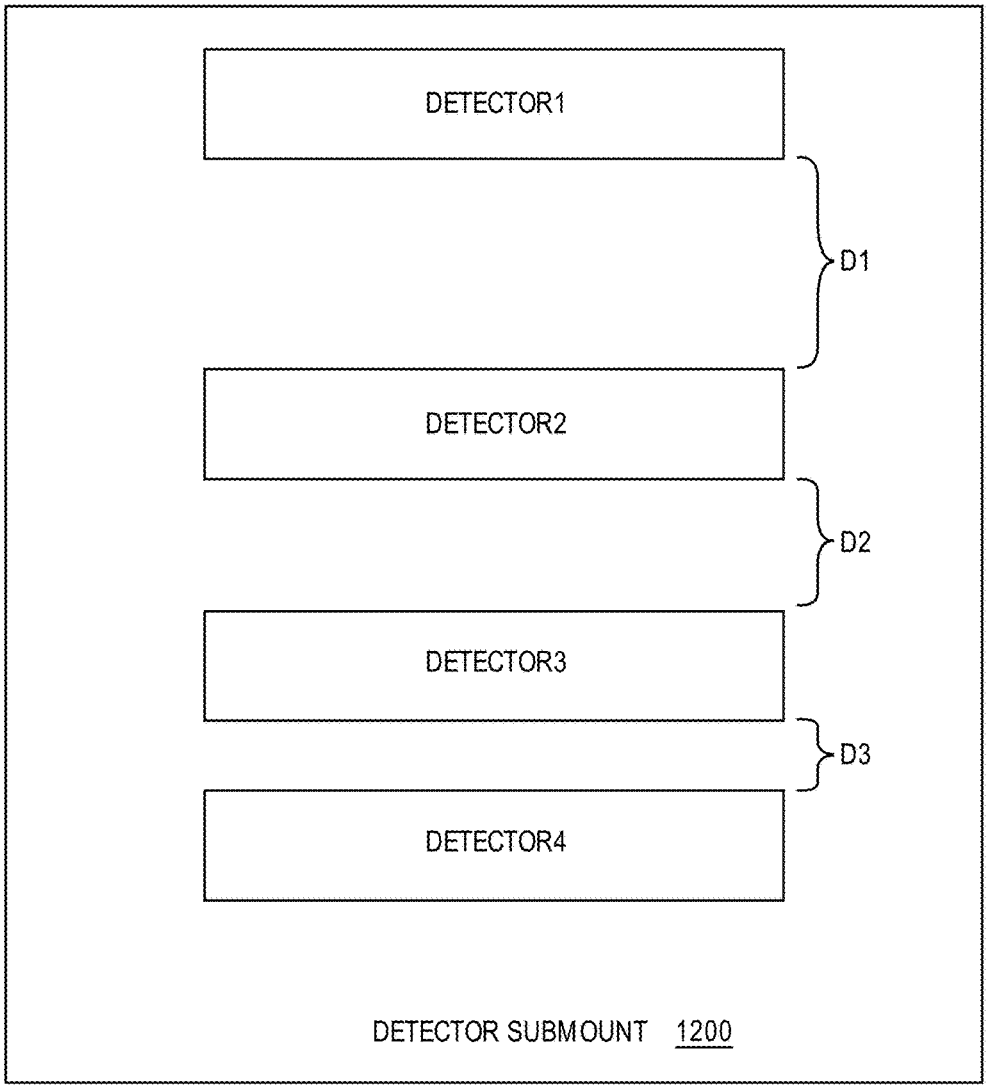

DETECTOR1
D1
DETECTOR2
D2
DETECTOR3
D3
DETECTOR4
DETECTOR SUBMOUNT 1200

FIG. 12C

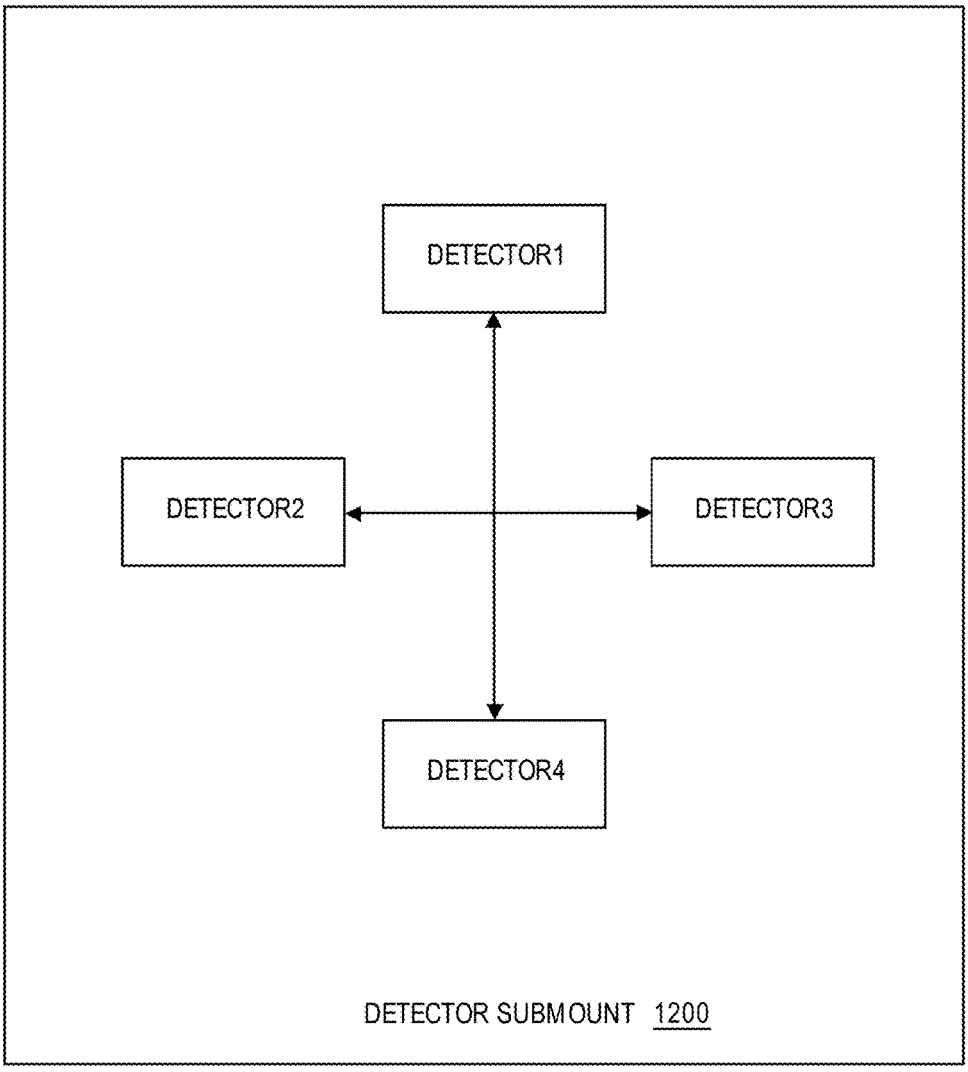

DETECTOR1
DETECTOR2
DETECTOR3
DETECTOR4
DETECTOR SUBMOUNT 1200

FIG. 12D

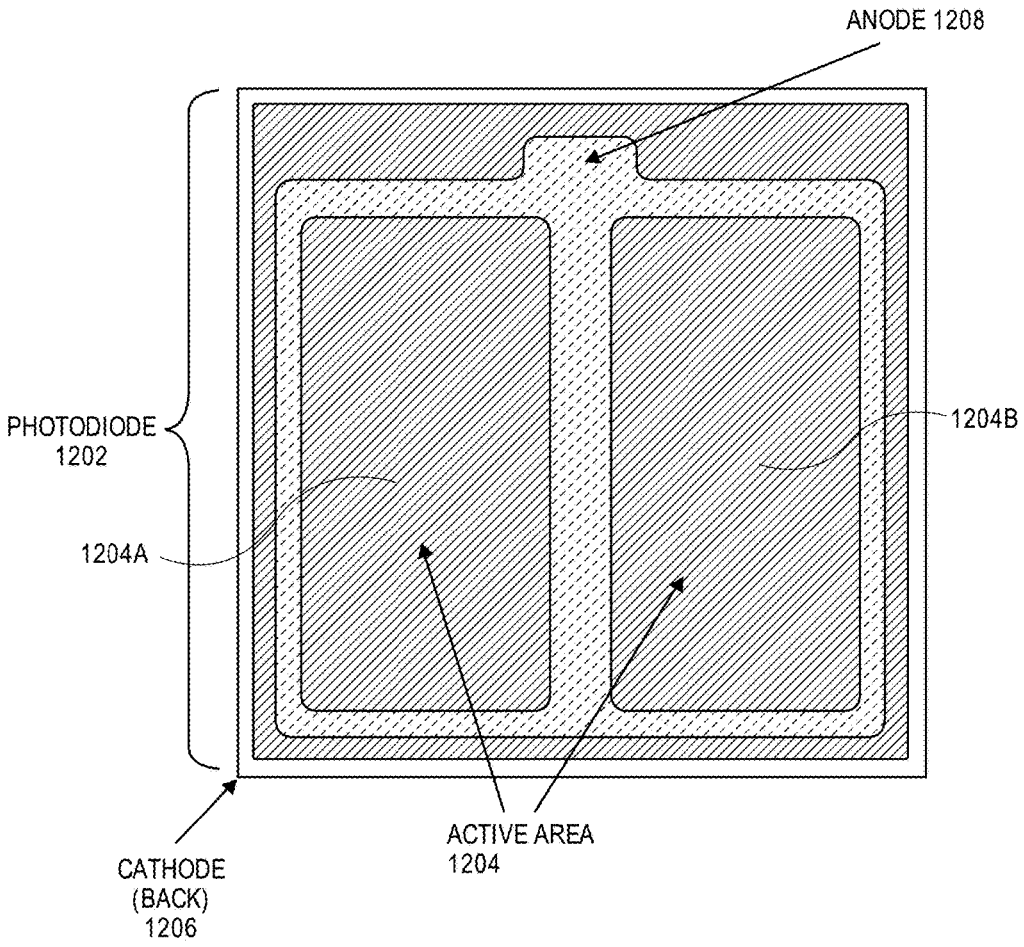

ANODE 1208
PHOTODIODE
1202
1204A
1204B
ACTIVE AREA
1204
CATHODE
(BACK)
1206

FIG. 12E

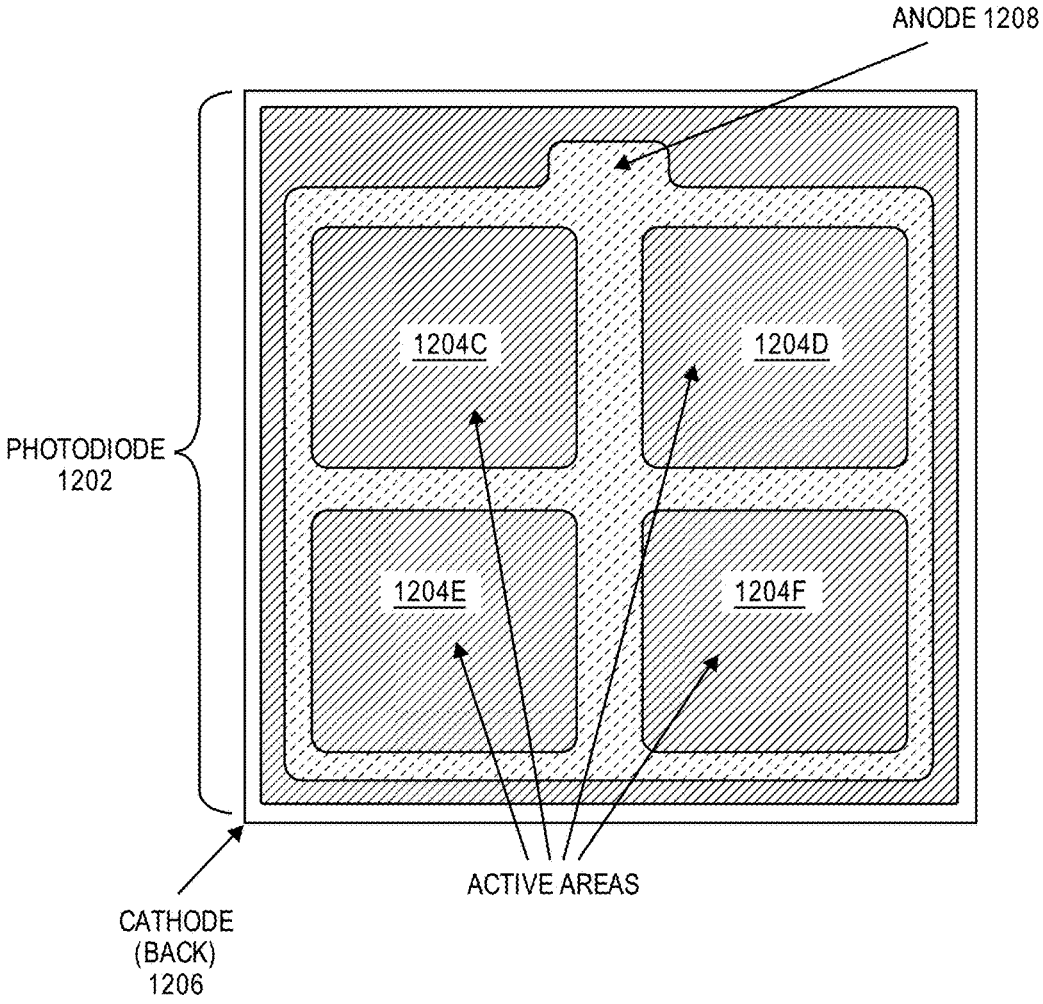

ANODE 1208
1204C
1204D
1204E
1204F
PHOTODIODE
1202
ACTIVE AREAS
CATHODE
(BACK)
1206

FIG. 12F

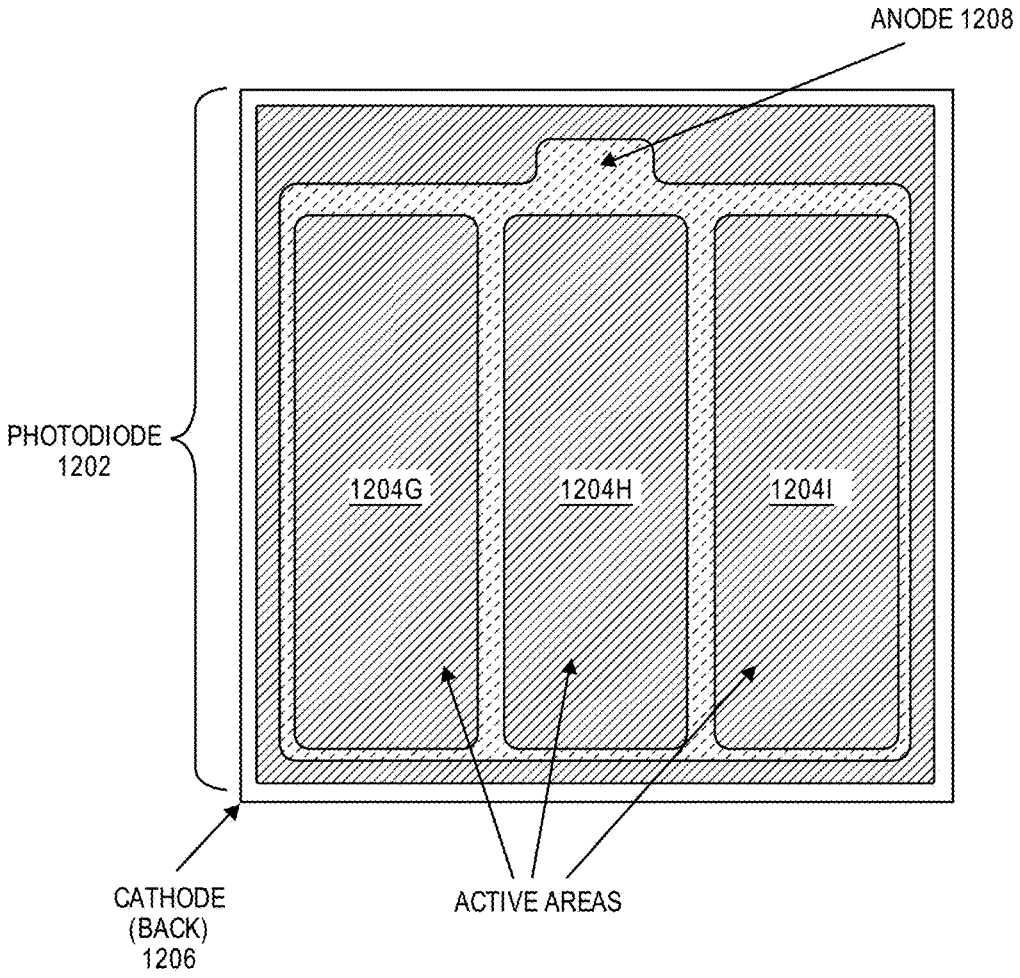

ANODE 1208
PHOTODIODE
1202
1204G
1204H
1204I
CATHODE
(BACK)
1206
ACTIVE AREAS

FIG. 12G

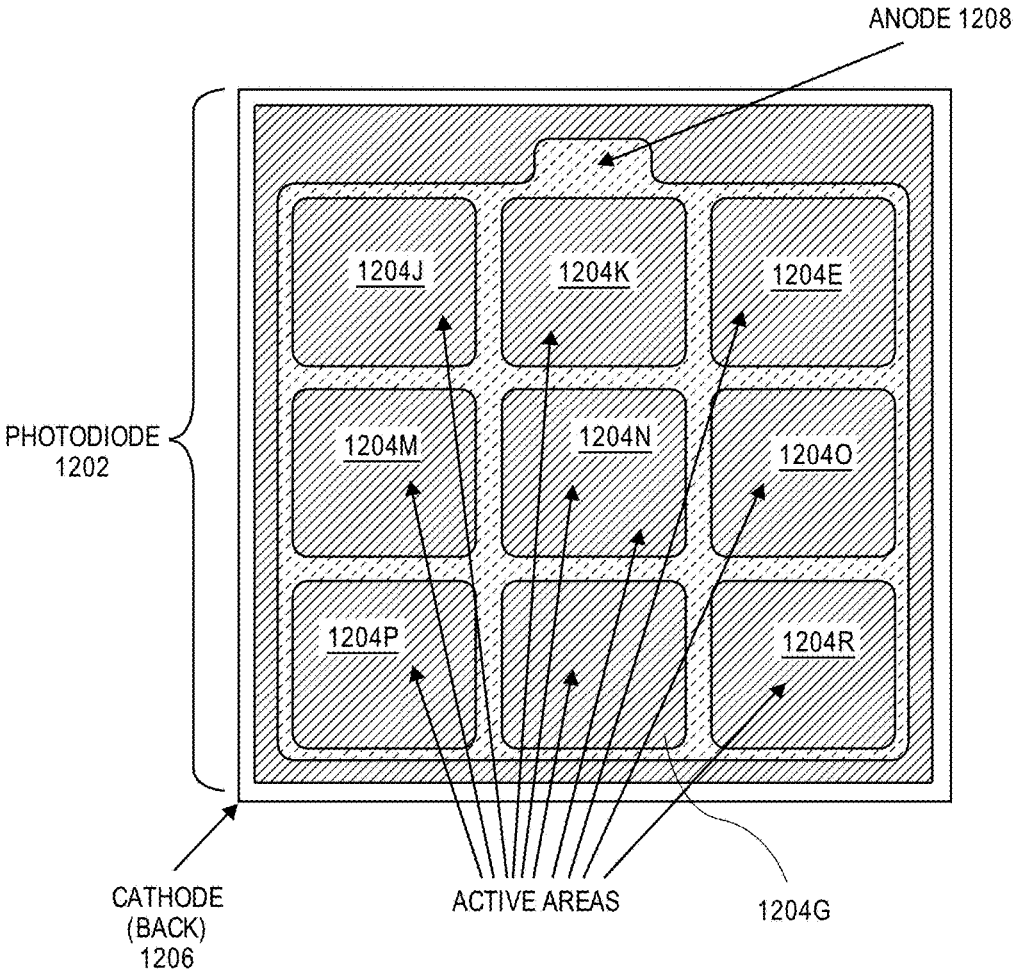

ANODE 1208
1204J
1204K
1204E
1204M
1204N
1204O
1204P
1204R
PHOTODIODE
1202
CATHODE
(BACK)
1206
ACTIVE AREAS
1204G

FIG. 12H

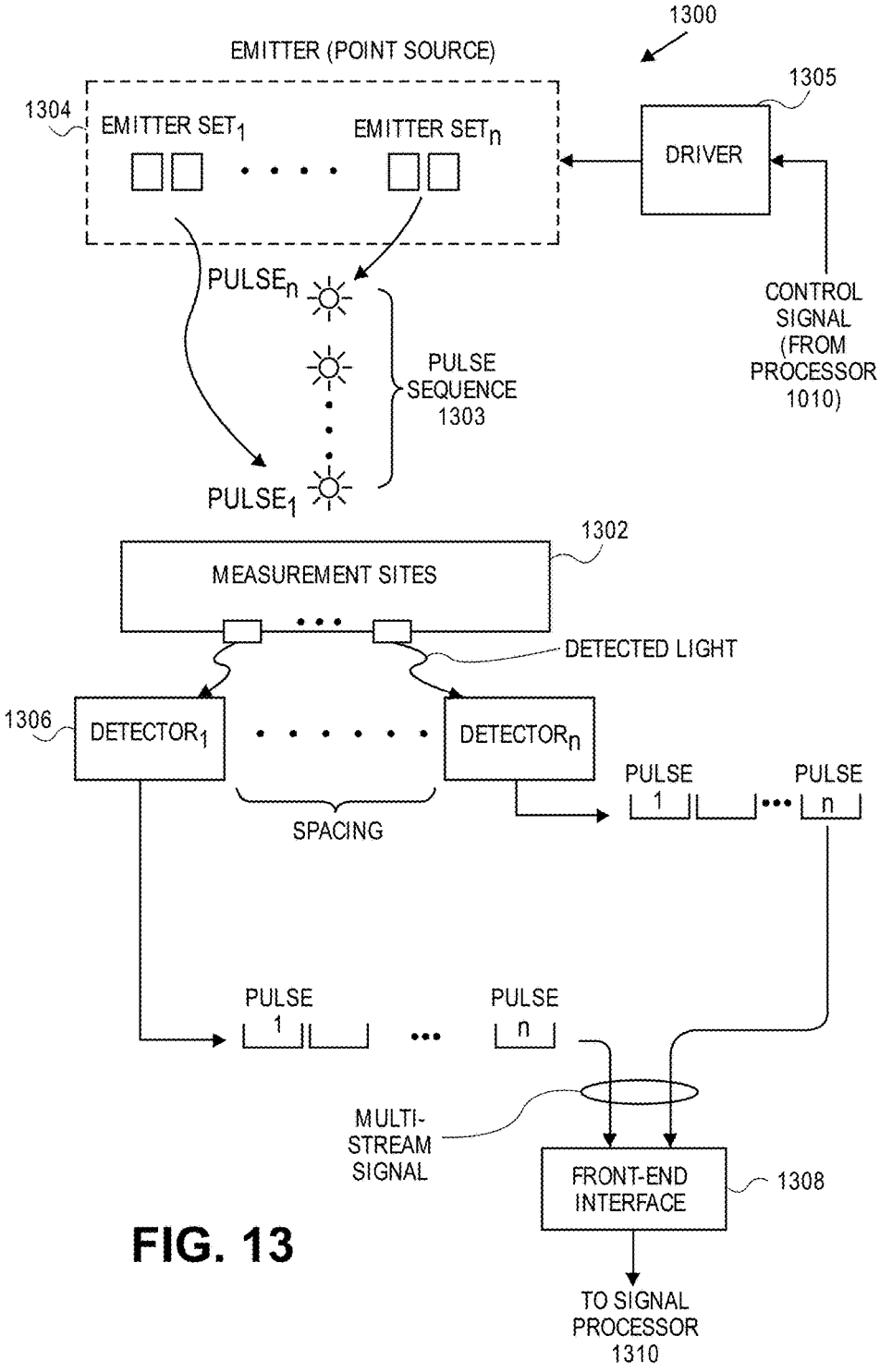

1300
EMITTER (POINT SOURCE)
1305
1304
EMITTER SET1
EMITTER SETn
DRIVER
PULSEn
PULSE SEQUENCE 1303
CONTROL SIGNAL (FROM PROCESSOR 1010)
PULSE1
1302
MEASUREMENT SITES
DETECTED LIGHT
1306
DETECTOR1
DETECTORn
PULSE 1
PULSE n
SPACING
PULSE 1
PULSE n
MULTI-STREAM SIGNAL
FRONT-END INTERFACE
1308
TO SIGNAL PROCESSOR 1310

FIG. 13

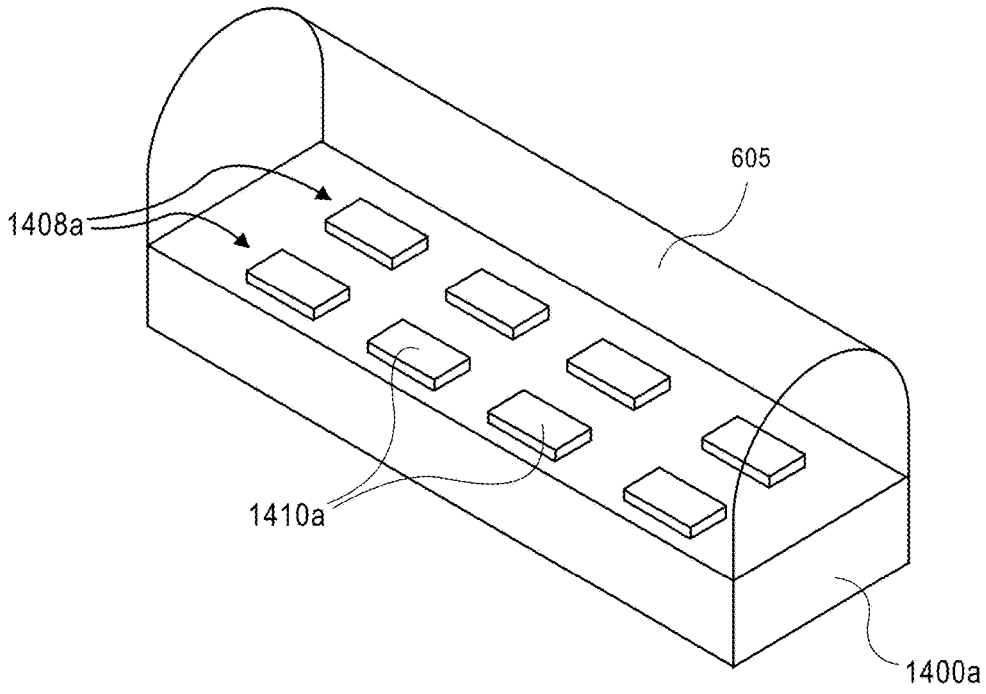


FIG. 14A

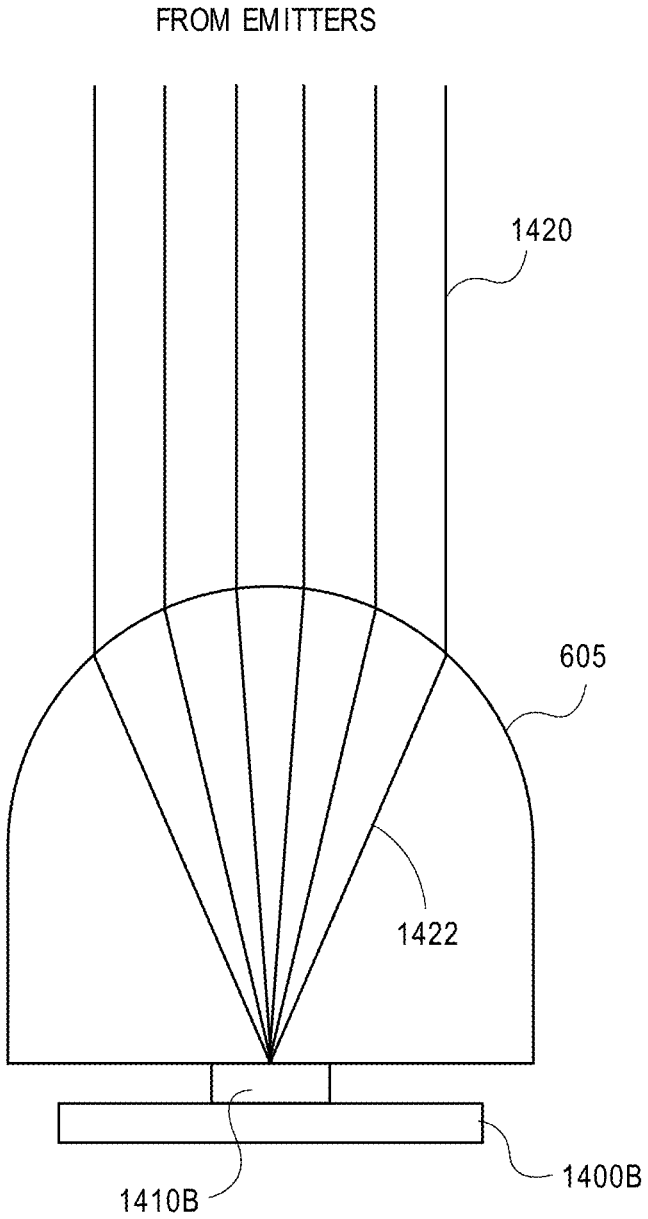


FIG. 14B

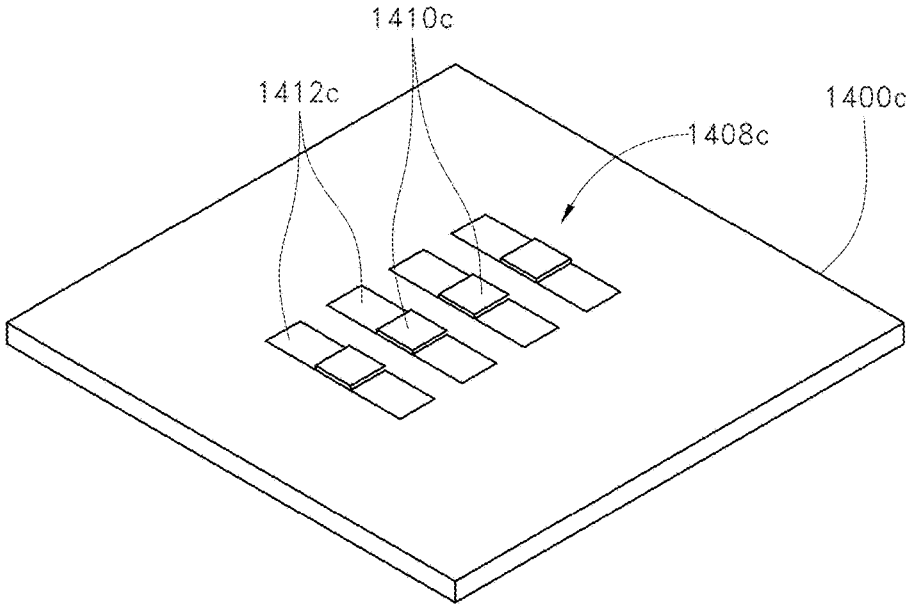


FIG. 14C

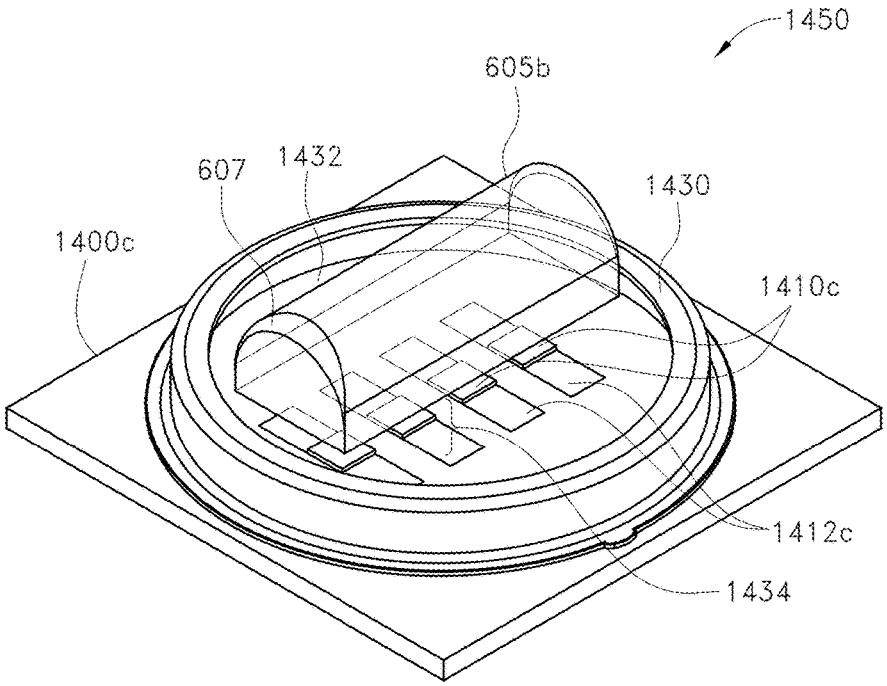


FIG. 14D

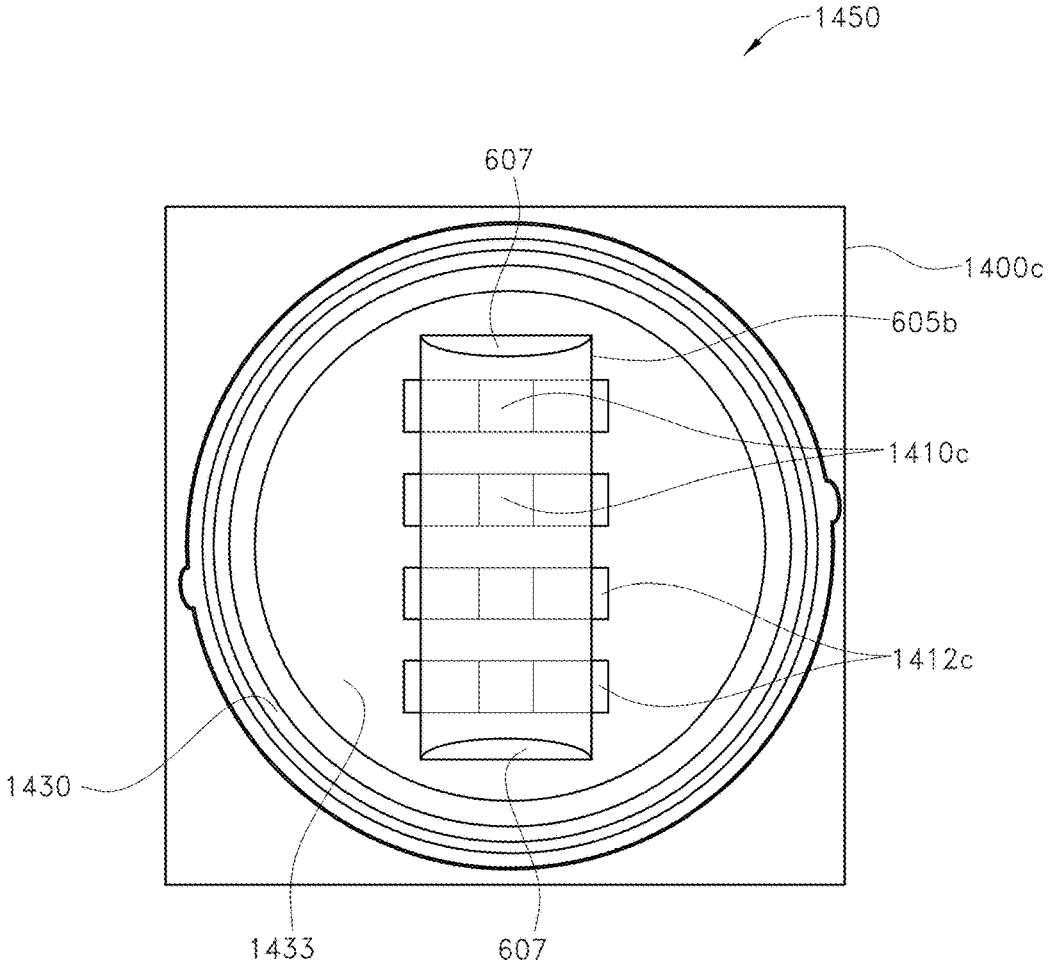


FIG. 14E

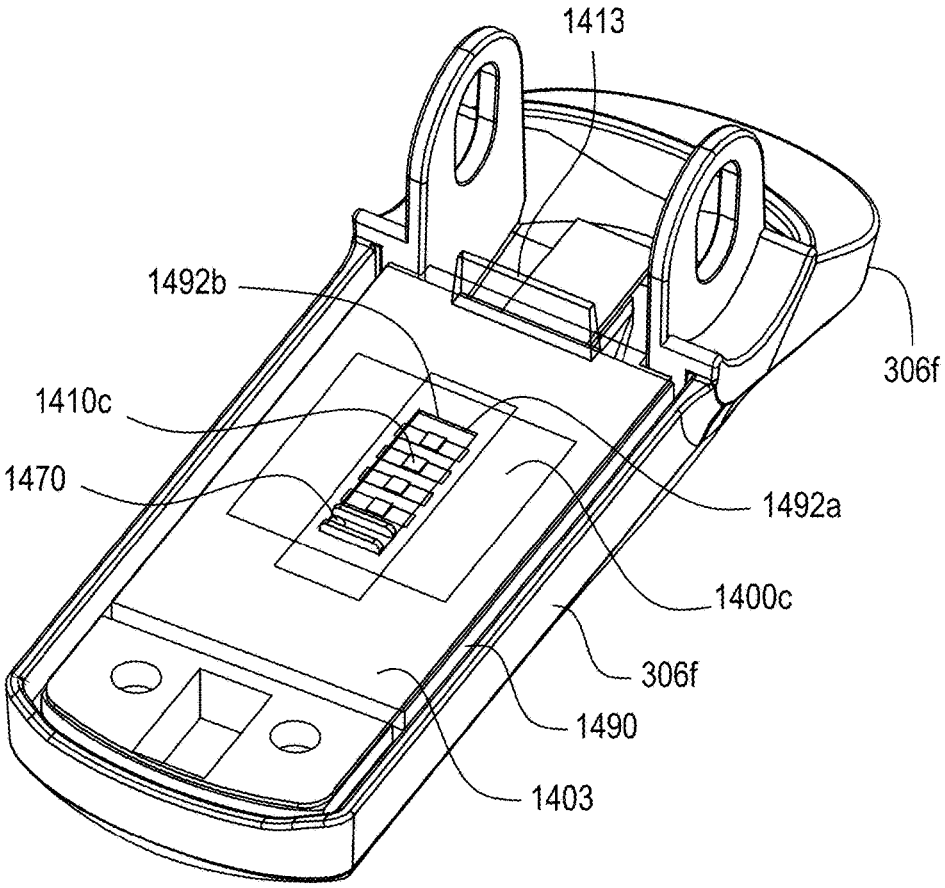


FIG. 14F

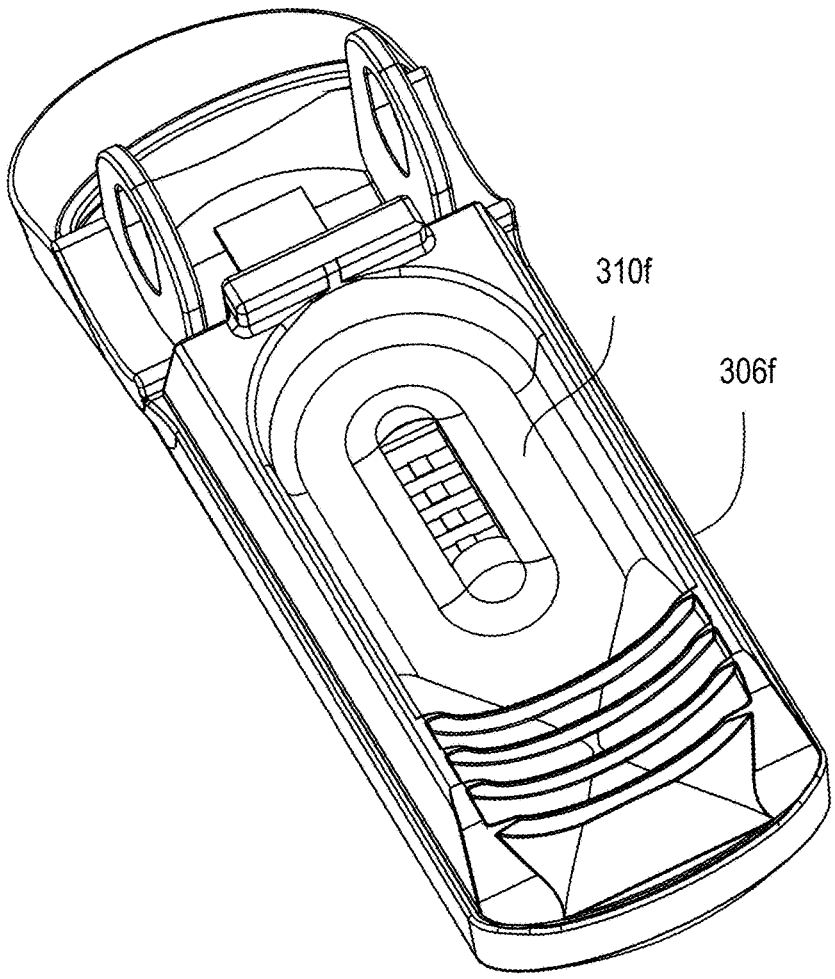


FIG. 14G

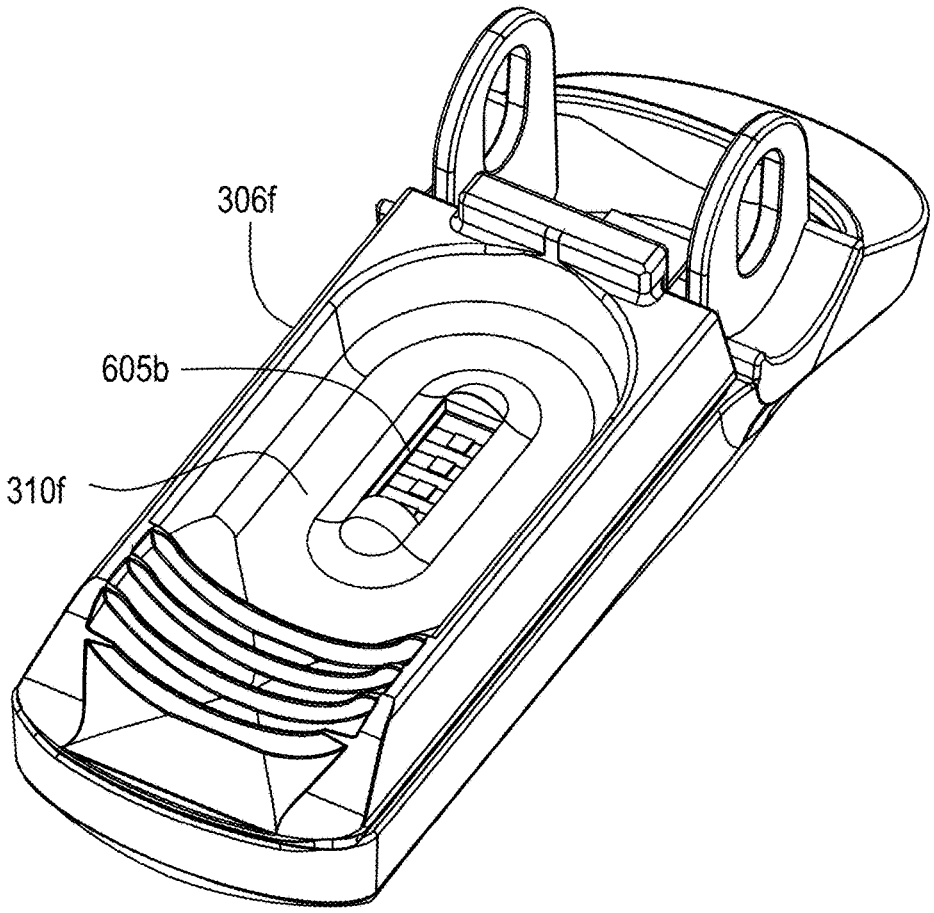


FIG. 14H

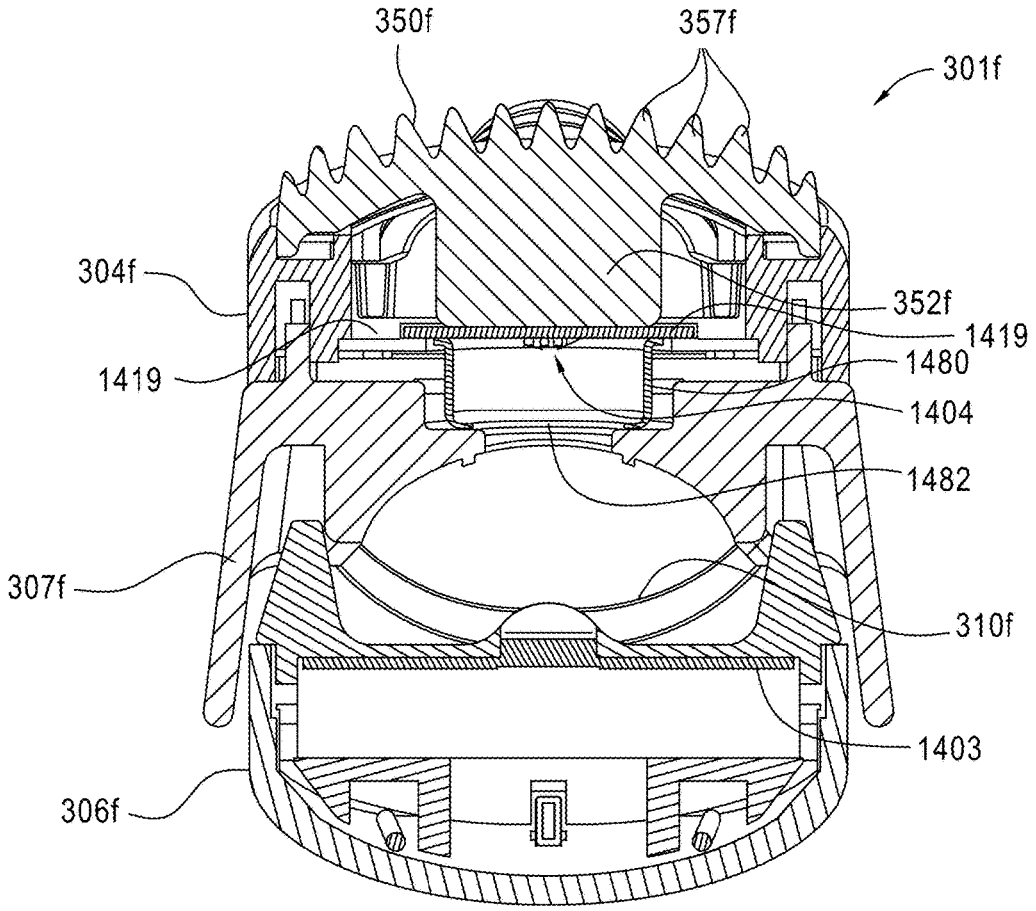

350f
357f
301f
304f
352f
1419
1419
1480
1404
1482
307f
310f
1403
306f

FIG. 14I

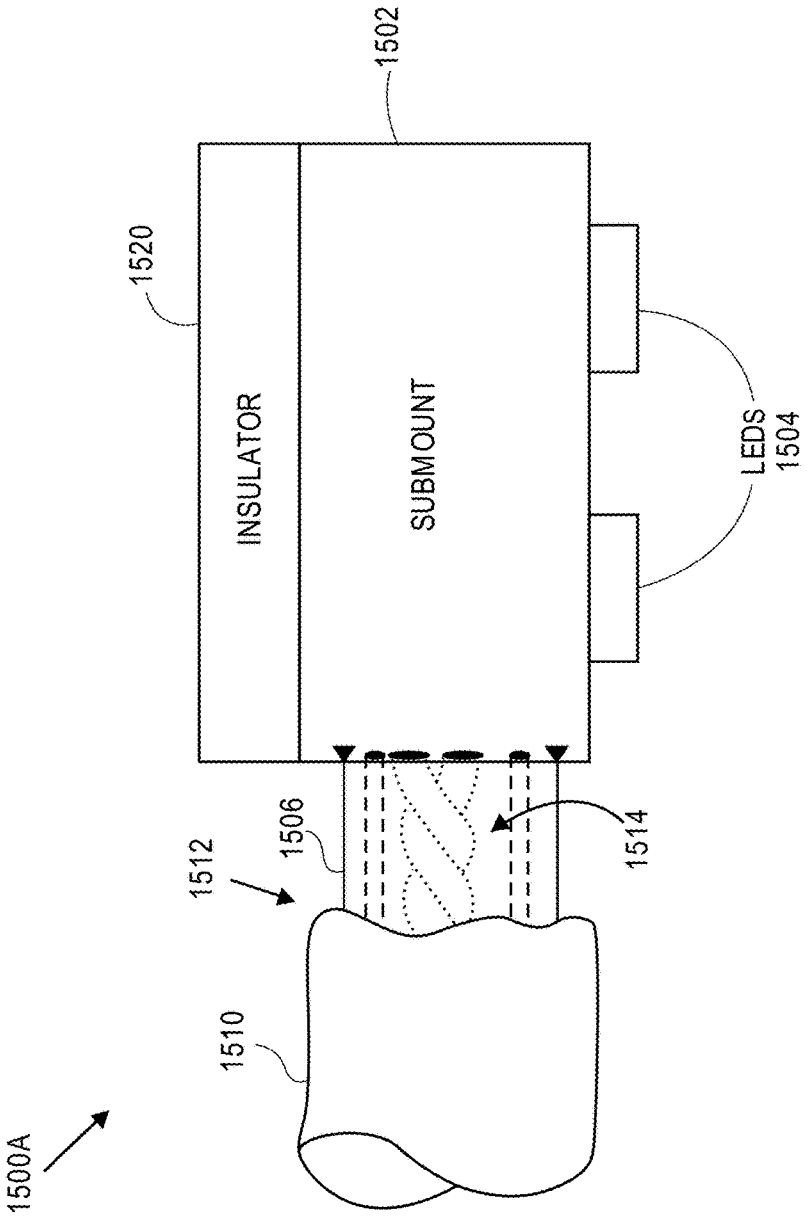

1502
1520
INSULATOR
SUBMOUNT
LEDS
1504
1506
1512
1514
1510
1500A

FIG. 15A

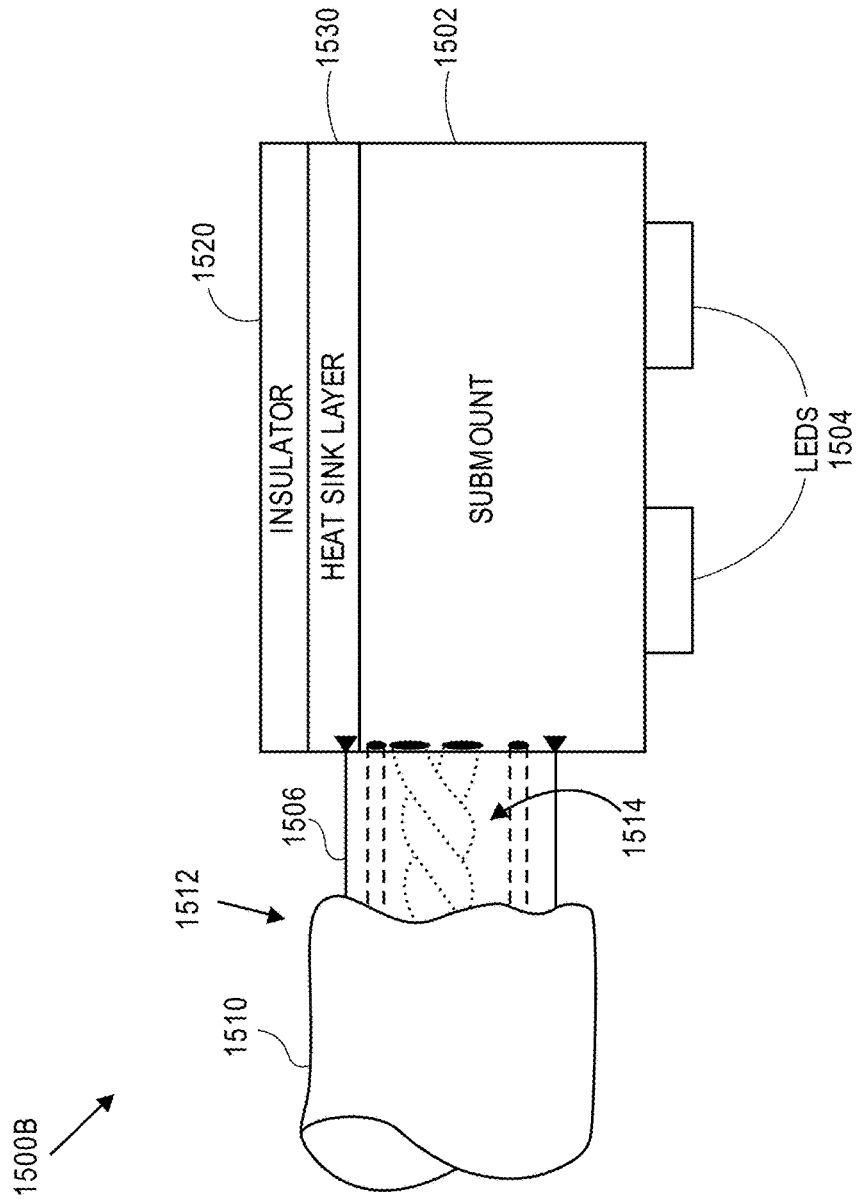

1530
1502
1520
INSULATOR
HEAT SINK LAYER
SUBMOUNT
LEDS
1504
1506
1514
1512
1510
1500B

FIG. 15B

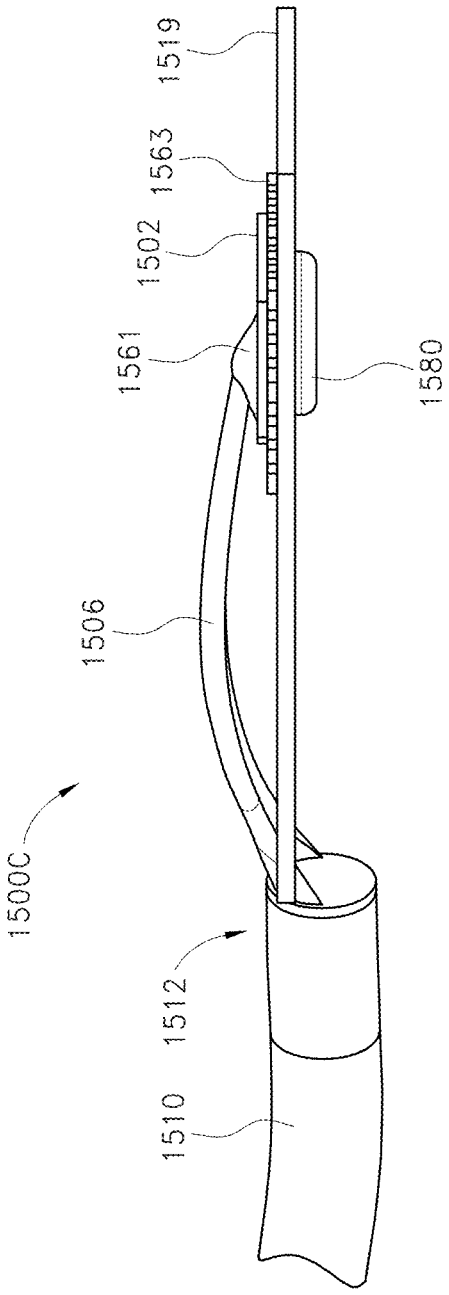


FIG. 15C

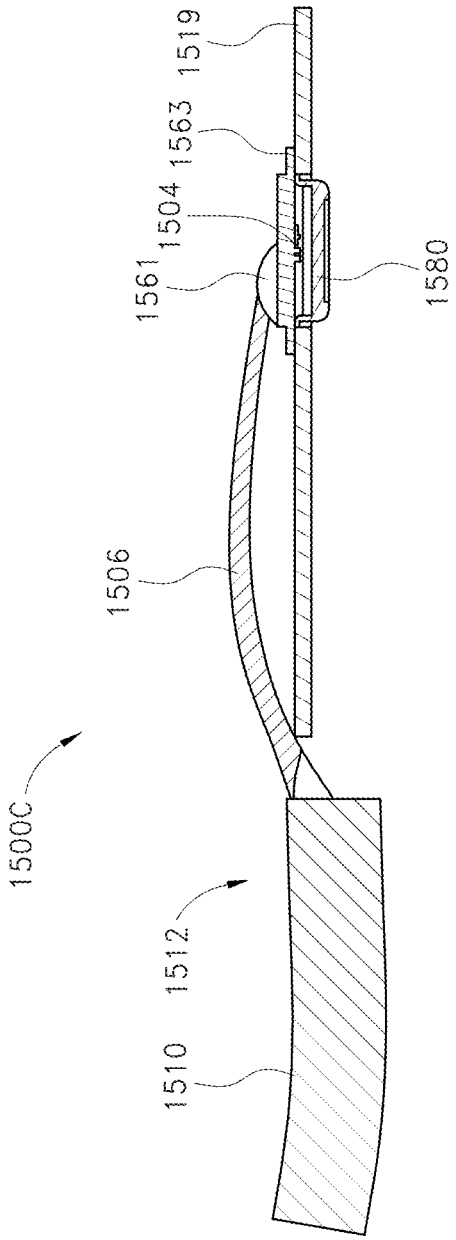


FIG. 15D

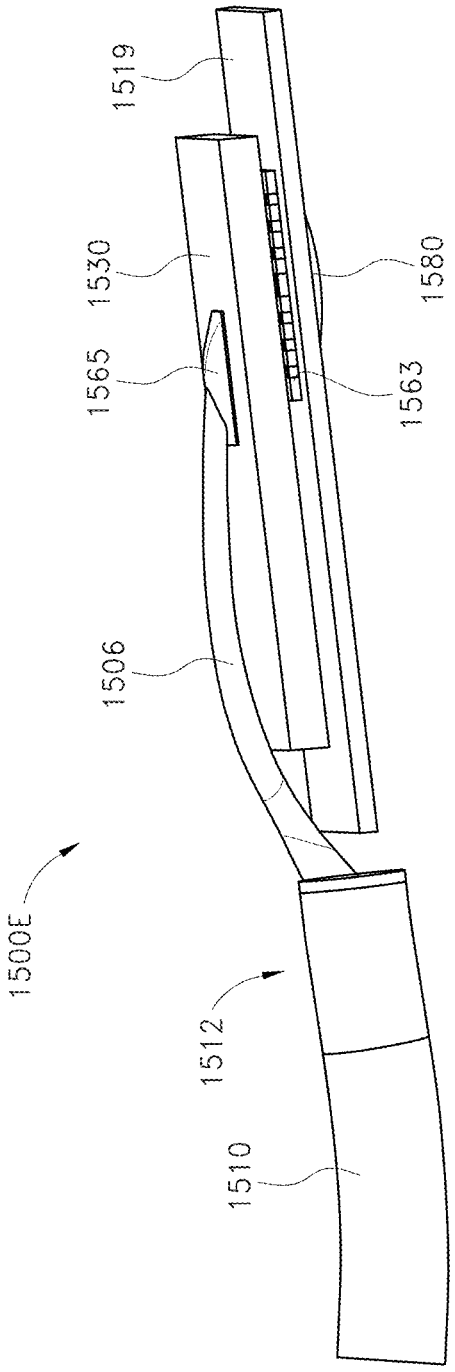


FIG. 15E

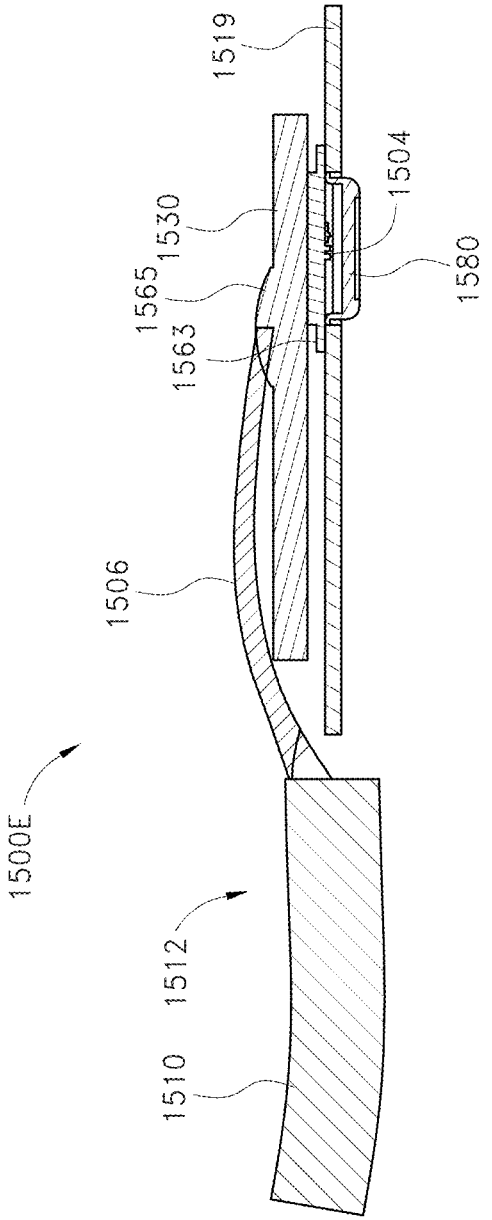


FIG. 15F

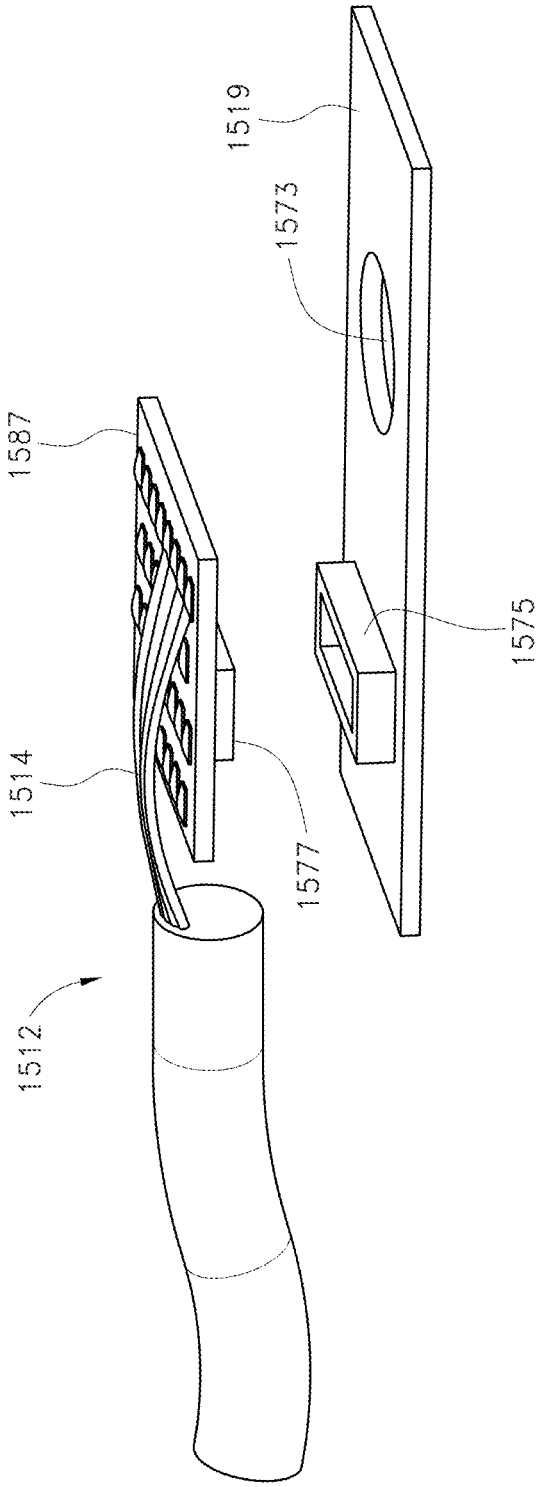


FIG. 15G

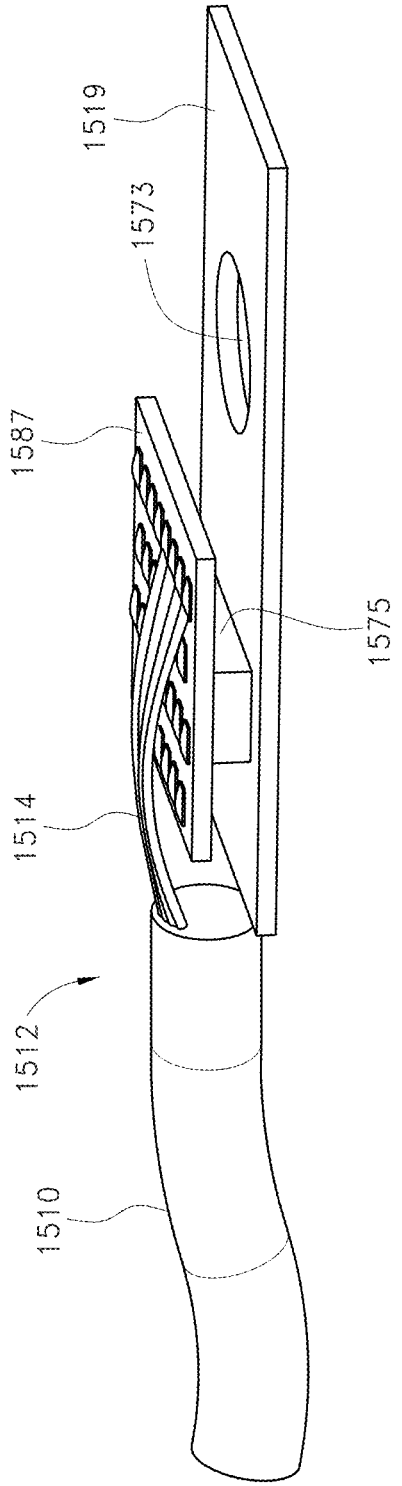


FIG. 15H

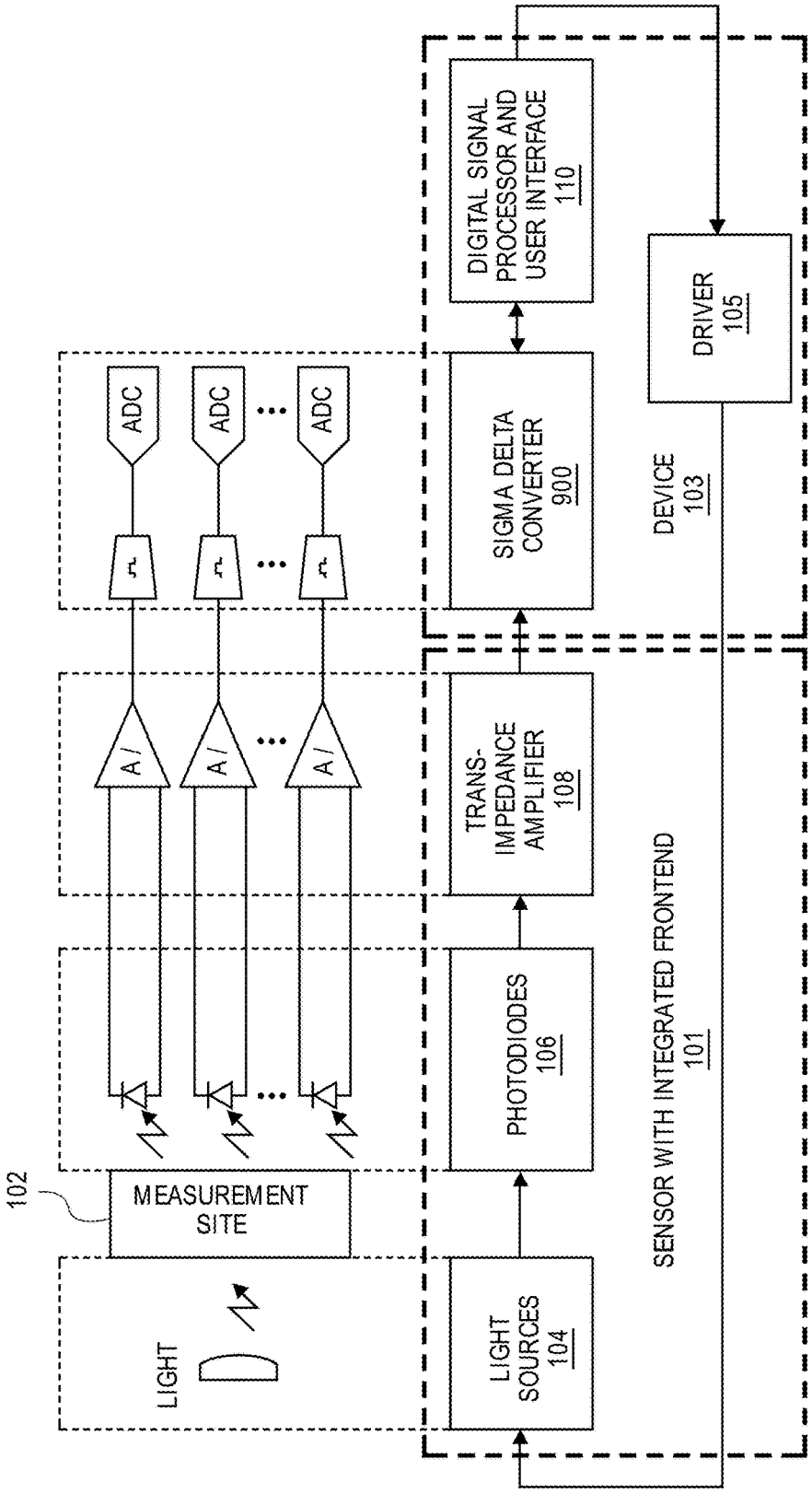

DIGITAL SIGNAL PROCESSOR AND USER INTERFACE 110
DRIVER 105
ADC
ADC
ADC
SIGMA DELTA CONVERTER 900
DEVICE 103
A/
A/
A/
TRANS-IMPEDANCE AMPLIFIER 108
PHOTODIODES 106
102
MEASUREMENT SITE
LIGHT
LIGHT SOURCES 104
SENSOR WITH INTEGRATED FRONTEND 101

FIG. 15I

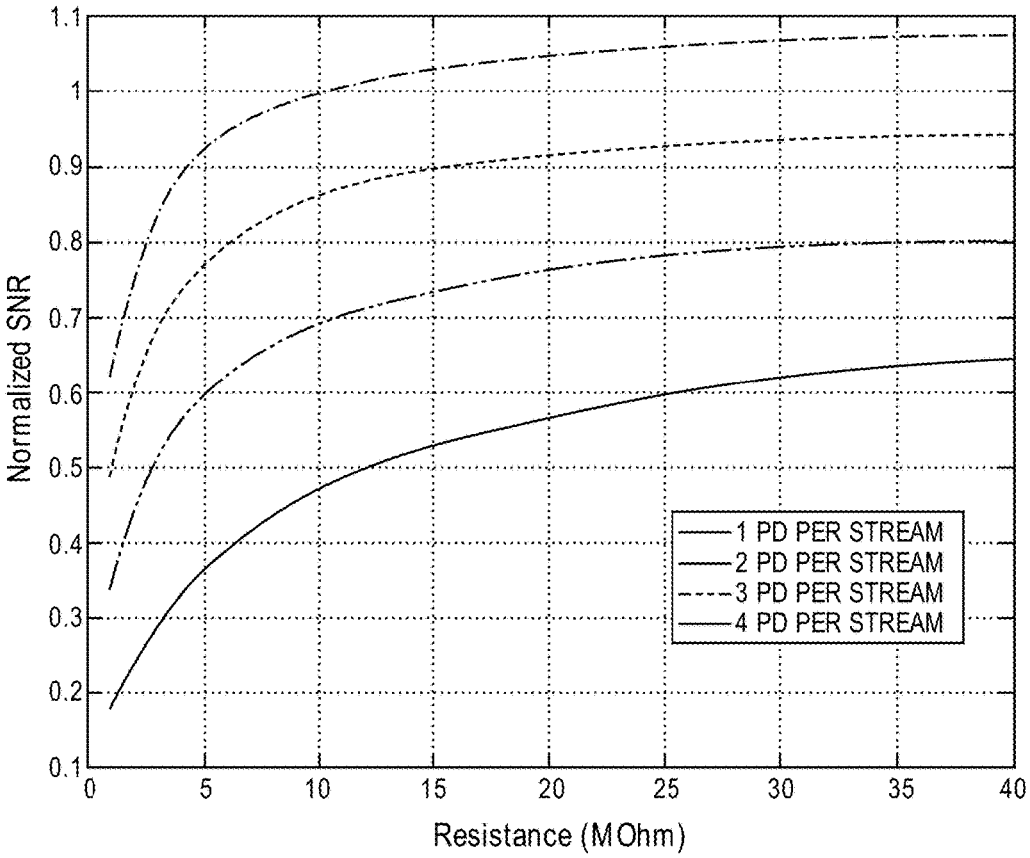

Normalized SNR
Resistance (MOhm)
1 PD PER STREAM
2 PD PER STREAM
3 PD PER STREAM
4 PD PER STREAM
1.1
1
0.9
0.8
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
5
10
15
20
25
30
35
40

FIG. 15J

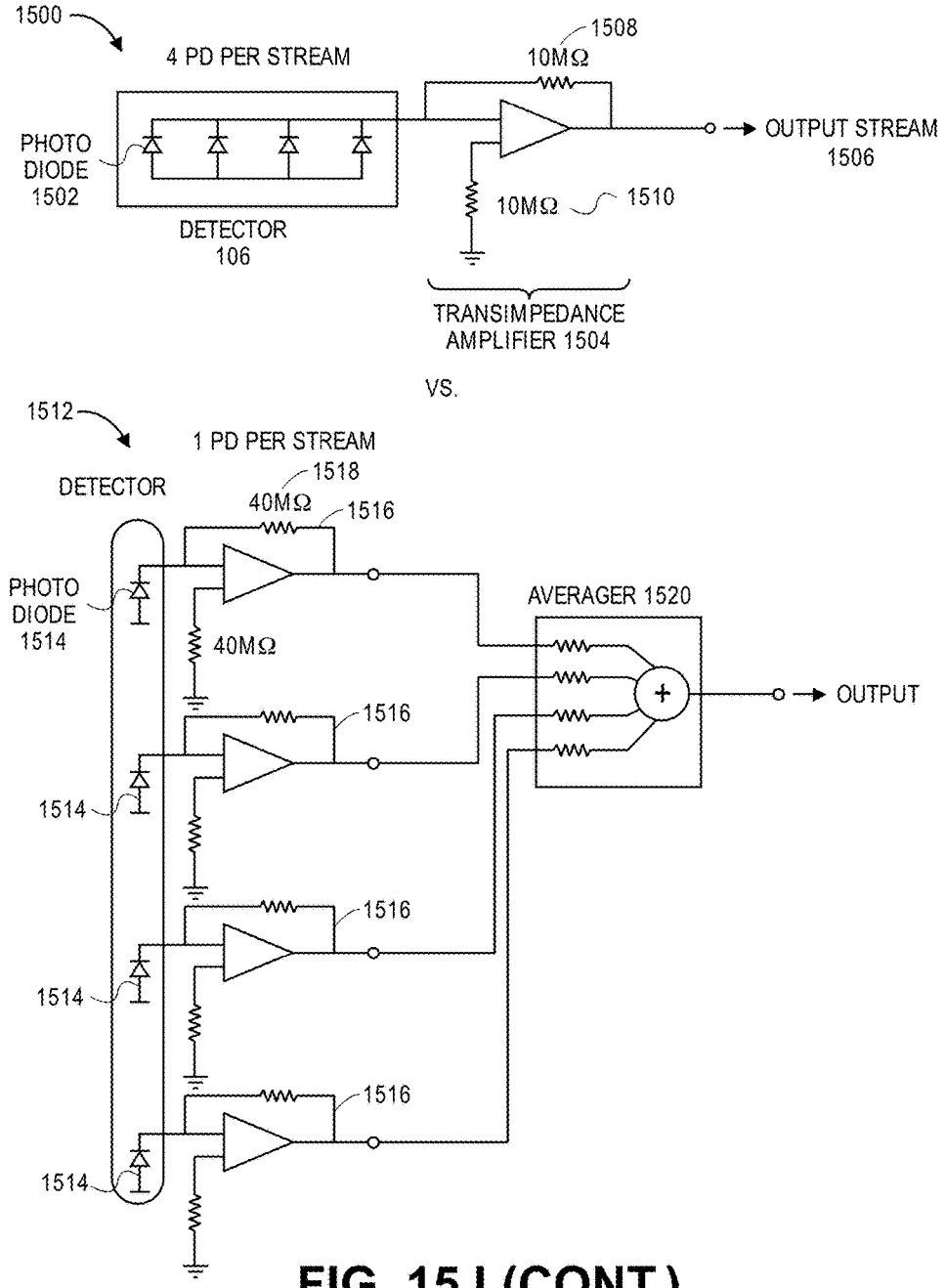
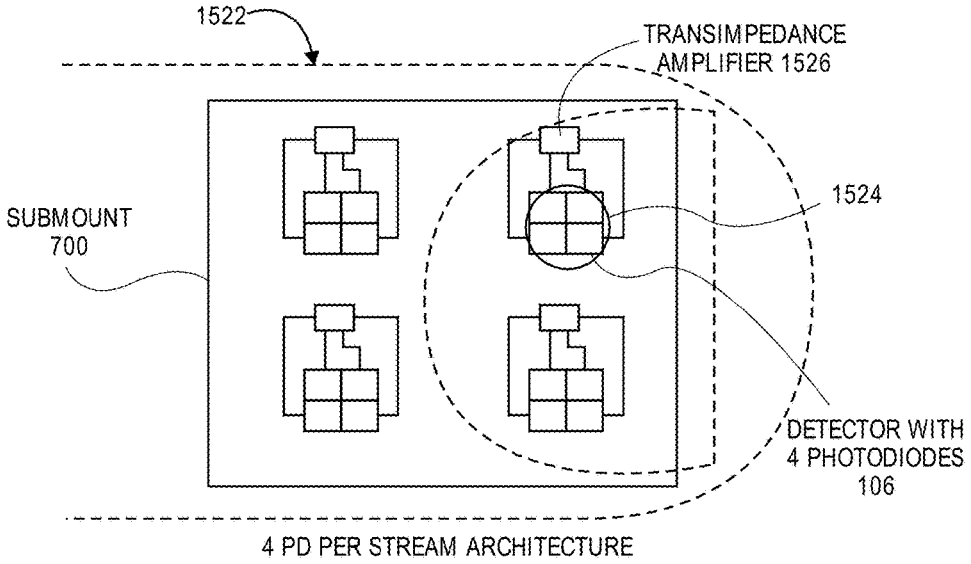

1500
4 PD PER STREAM
1508
10MΩ
PHOTO DIODE 1502
OUTPUT STREAM 1506
10MΩ
1510
DETECTOR 106
TRANSIMPEDANCE AMPLIFIER 1504
VS.
1512
1 PD PER STREAM
DETECTOR
1518
40MΩ
1516
PHOTO DIODE 1514
AVERAGER 1520
40MΩ
OUTPUT
1514
1516
1514
1516
1514
1516

FIG. 15J (CONT.)


1522
TRANSIMPEDANCE AMPLIFIER 1526
SUBMOUNT 700
1524
DETECTOR WITH 4 PHOTODIODES 106
4 PD PER STREAM ARCHITECTURE

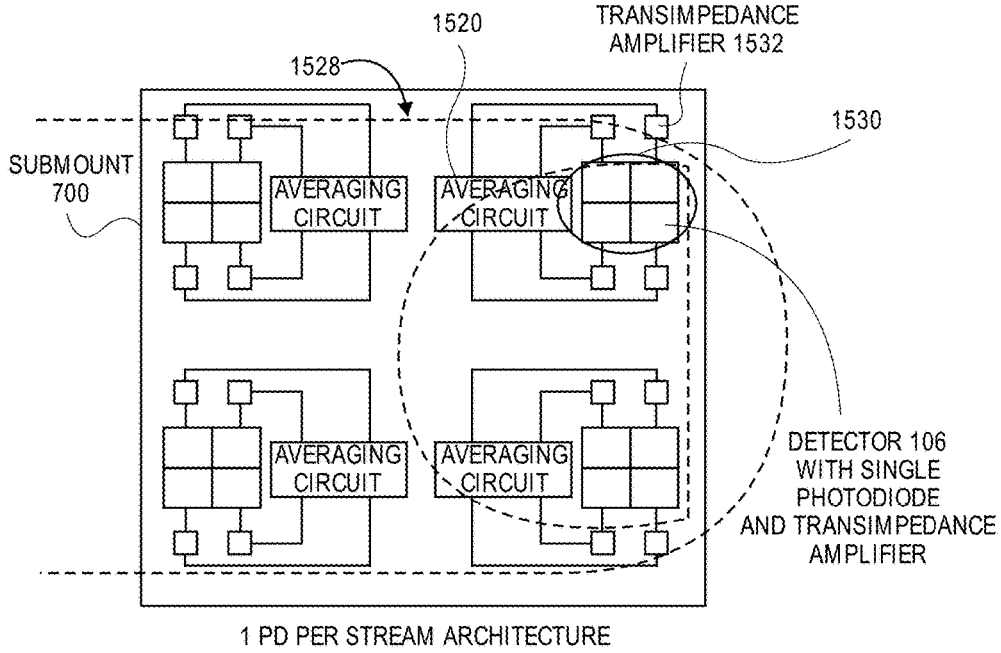

1520
TRANSIMPEDANCE AMPLIFIER 1532
1528
1530
SUBMOUNT 700
AVERAGING CIRCUIT
AVERAGING CIRCUIT
AVERAGING CIRCUIT
AVERAGING CIRCUIT
DETECTOR 106 WITH SINGLE PHOTODIODE AND TRANSIMPEDANCE AMPLIFIER
1 PD PER STREAM ARCHITECTURE

FIG. 15K

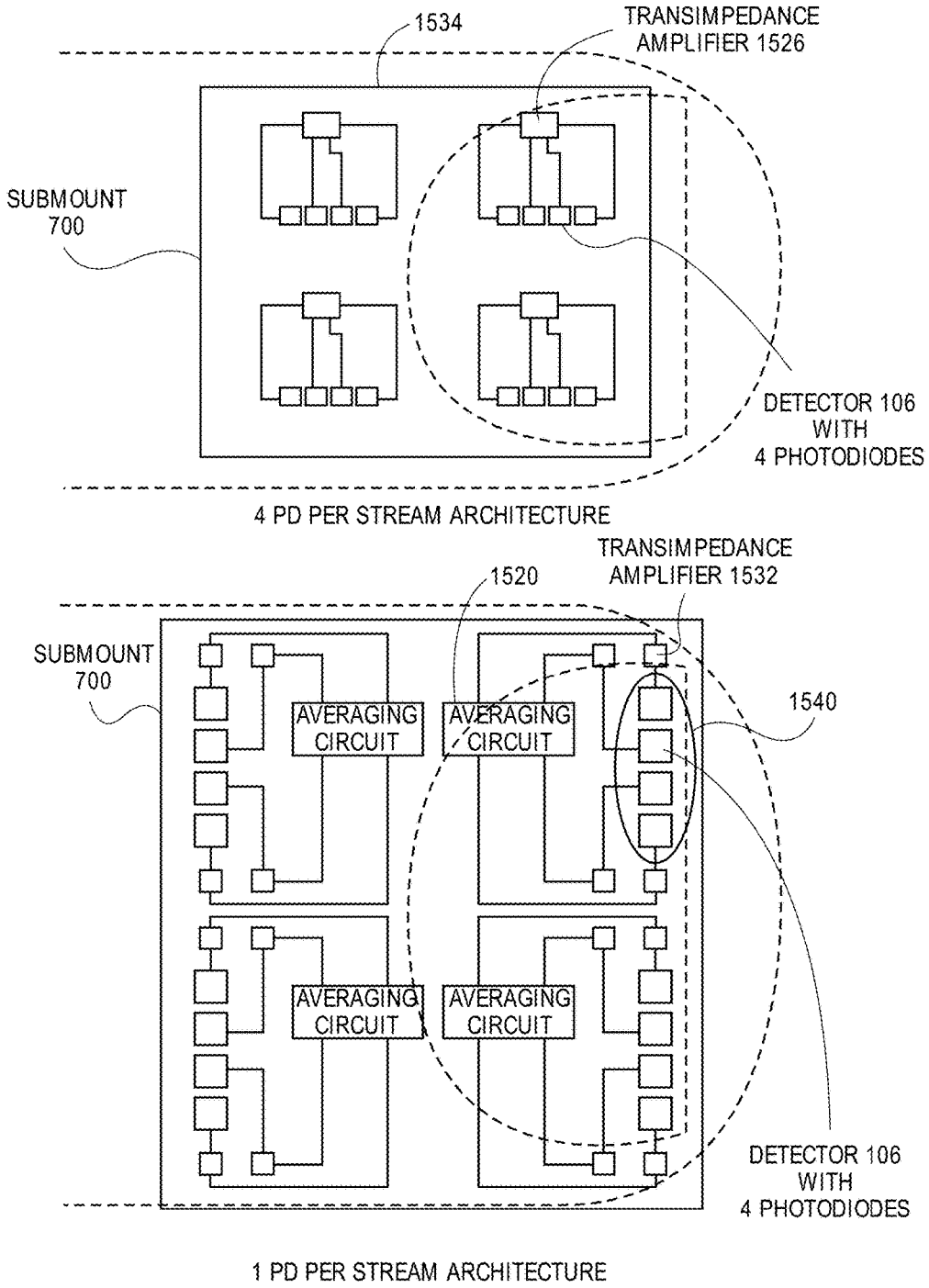

1534
TRANSIMPEDANCE AMPLIFIER 1526
SUBMOUNT 700
DETECTOR 106 WITH 4 PHOTODIODES
4 PD PER STREAM ARCHITECTURE
TRANSIMPEDANCE AMPLIFIER 1532
1520
SUBMOUNT 700
AVERAGING CIRCUIT
AVERAGING CIRCUIT
1540
AVERAGING CIRCUIT
AVERAGING CIRCUIT
DETECTOR 106 WITH 4 PHOTODIODES
1 PD PER STREAM ARCHITECTURE

FIG. 15K (CONT.)

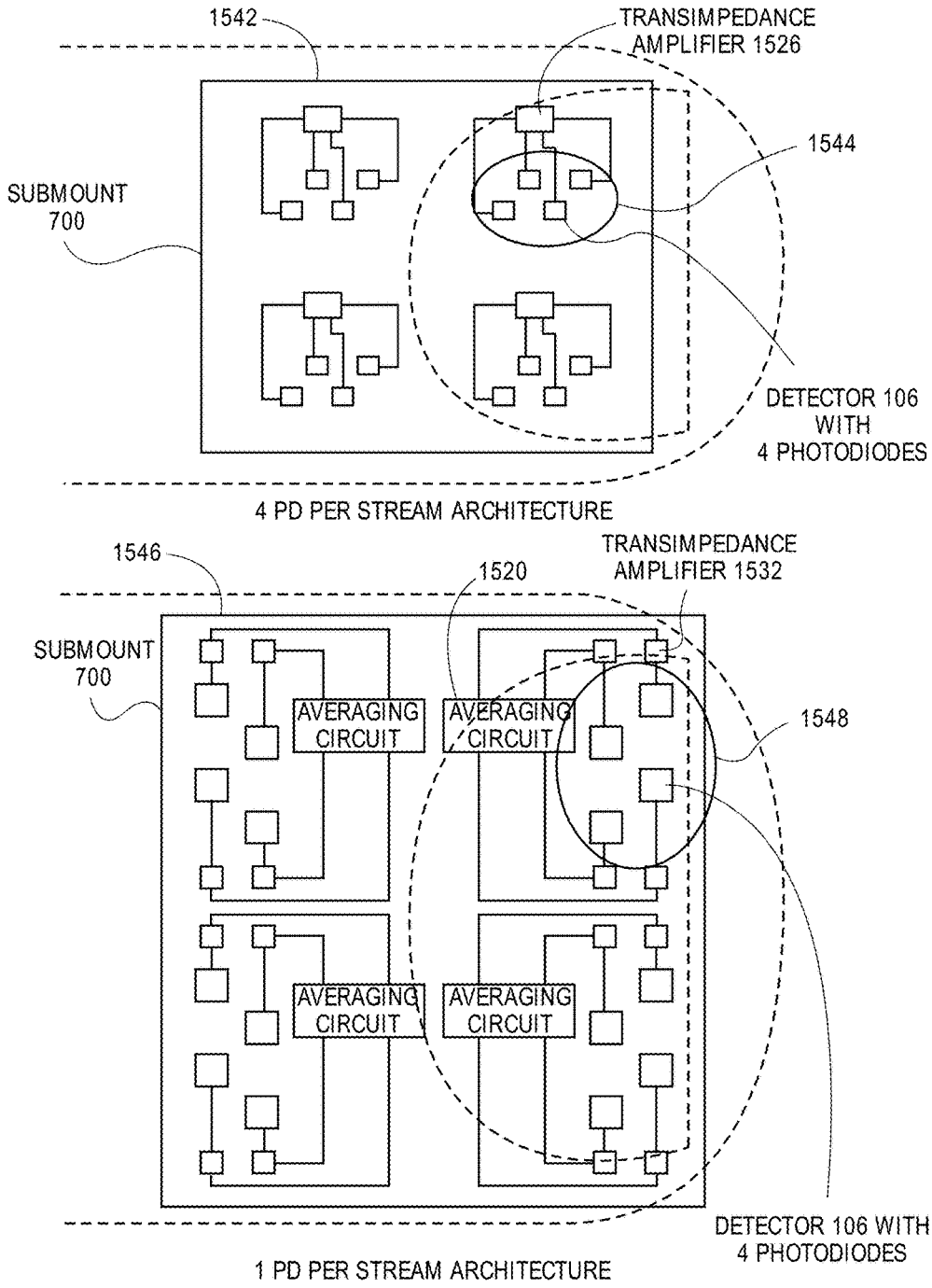

1542
TRANSIMPEDANCE AMPLIFIER 1526
1544
SUBMOUNT 700
DETECTOR 106 WITH 4 PHOTODIODES
4 PD PER STREAM ARCHITECTURE
1546
1520
TRANSIMPEDANCE AMPLIFIER 1532
SUBMOUNT 700
AVERAGING CIRCUIT
AVERAGING CIRCUIT
1548
AVERAGING CIRCUIT
AVERAGING CIRCUIT
DETECTOR 106 WITH 4 PHOTODIODES
1 PD PER STREAM ARCHITECTURE

FIG. 15K (CONT.)

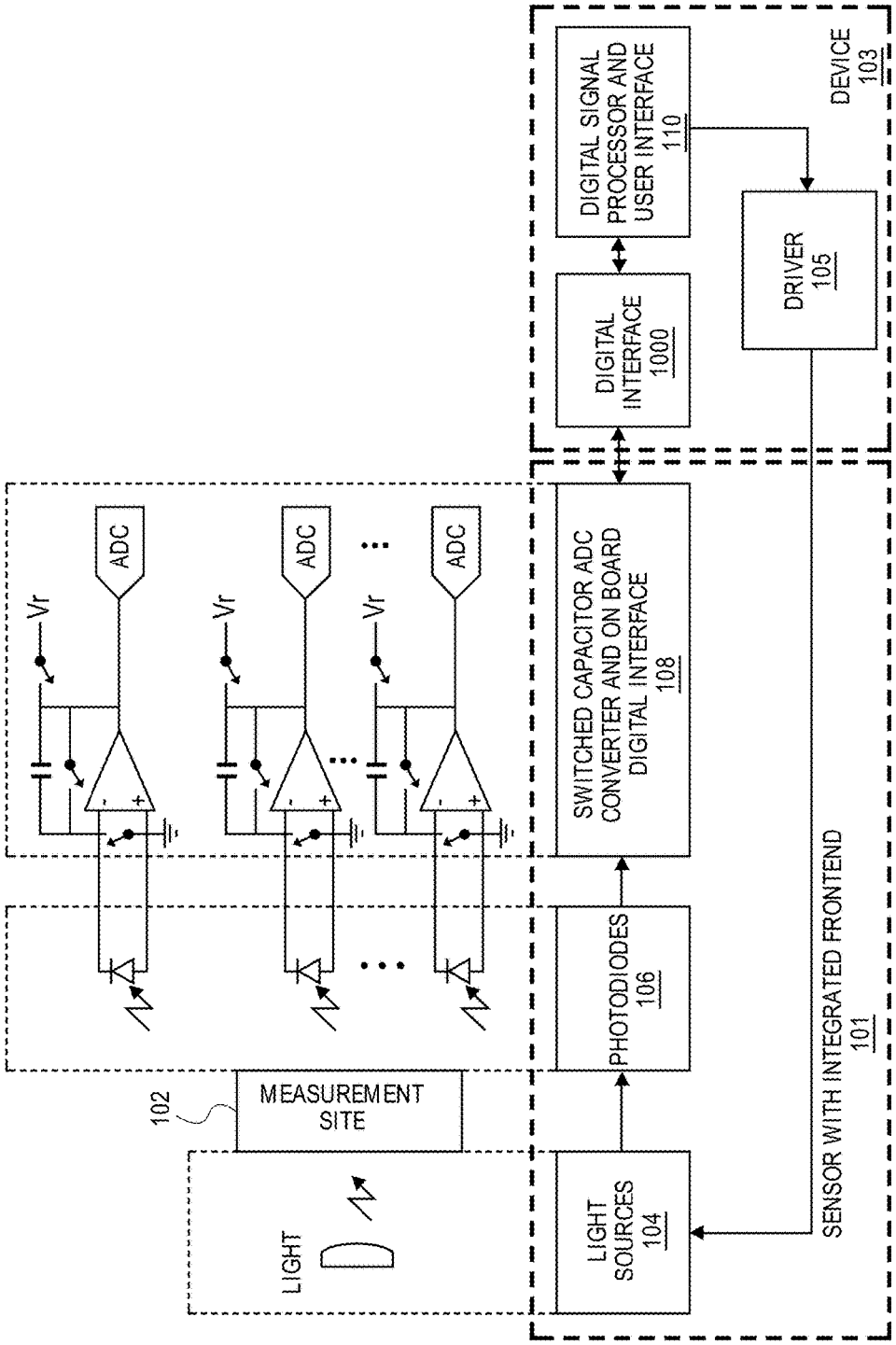

DEVICE
103
DIGITAL SIGNAL PROCESSOR AND USER INTERFACE
110
DRIVER
105
DIGITAL INTERFACE
1000
SWITCHED CAPACITOR ADC CONVERTER AND ON BOARD DIGITAL INTERFACE
108
PHOTODIODES
106
LIGHT SOURCES
104
SENSOR WITH INTEGRATED FRONTEND
101
MEASUREMENT SITE
102
LIGHT
ADC
Vr

FIG. 15L

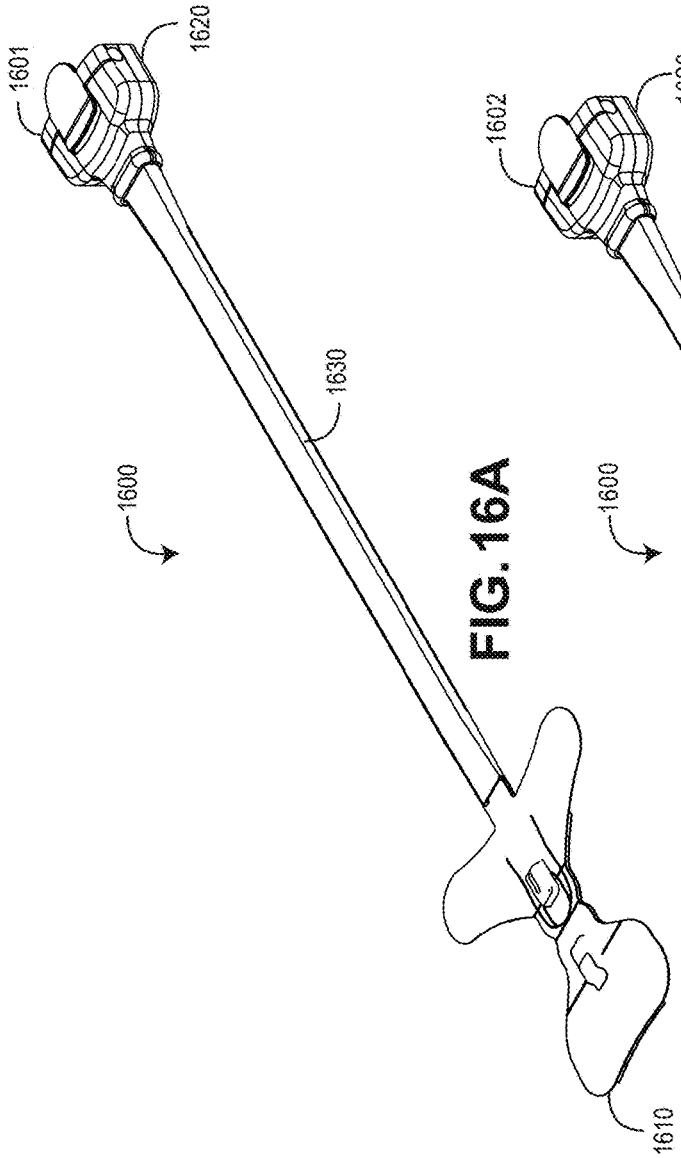


FIG. 16A

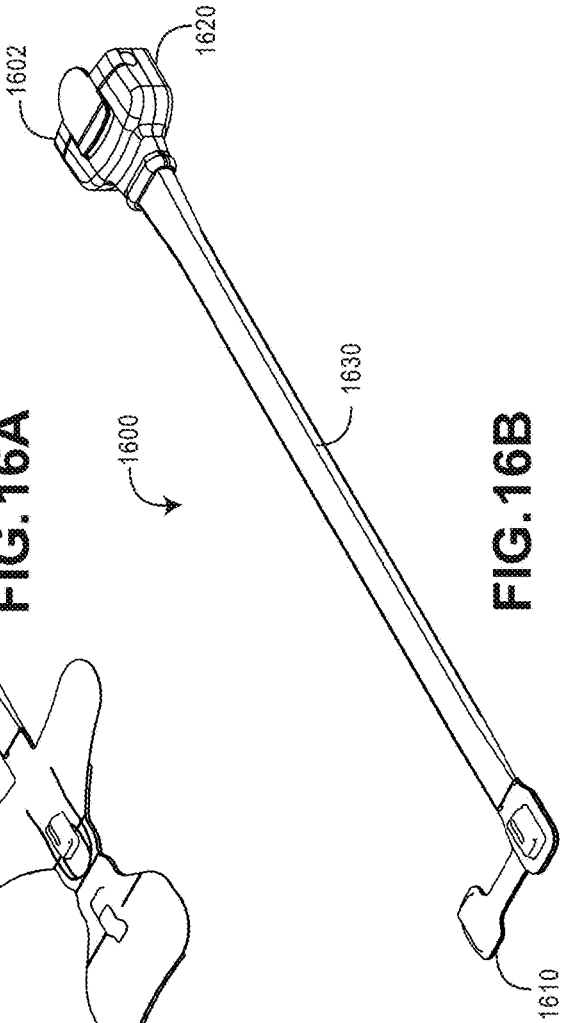


FIG. 16B

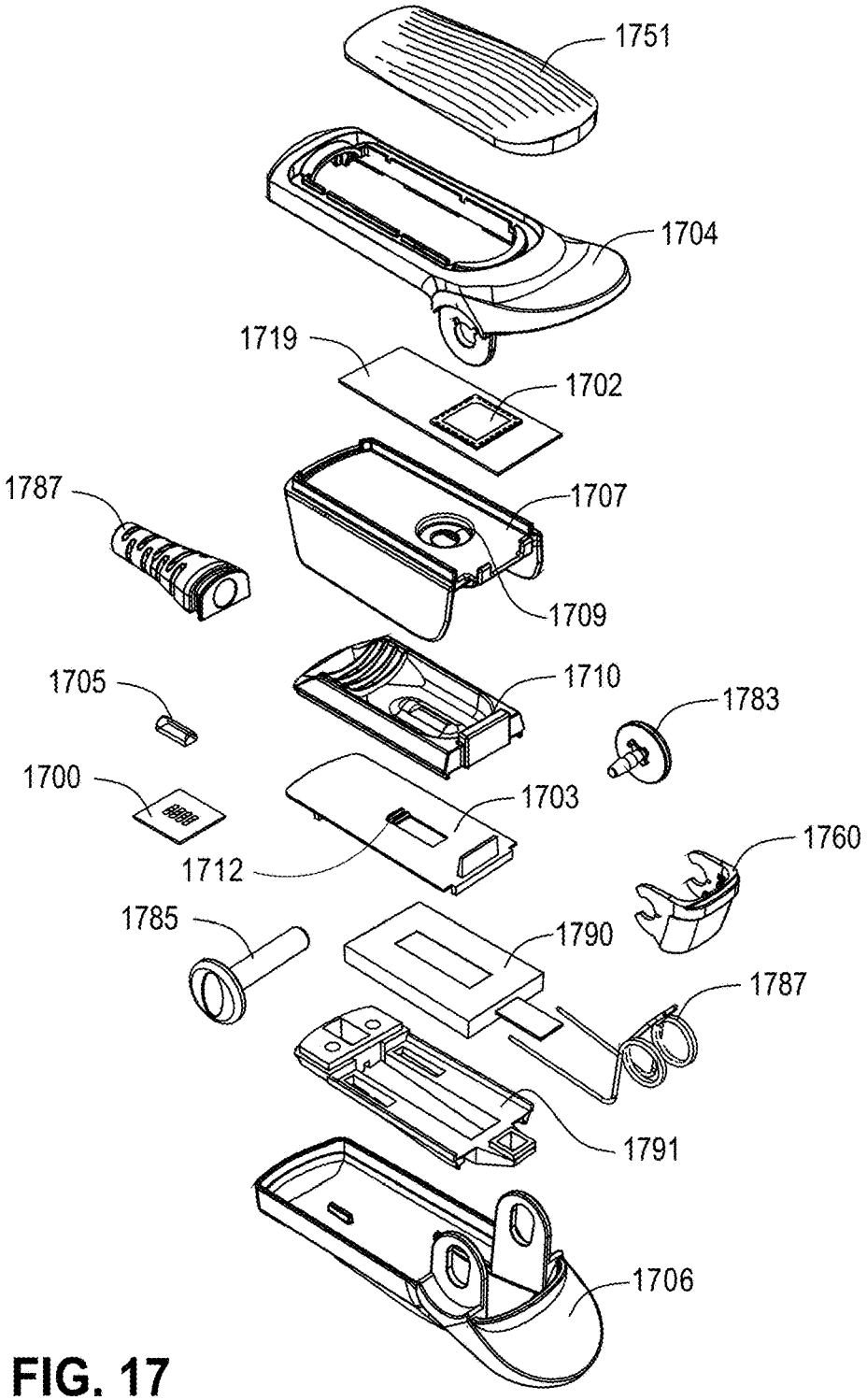

1751
1704
1719
1702
1787
1707
1709
1705
1710
1783
1700
1703
1760
1712
1785
1790
1787
1791
1706

FIG. 17

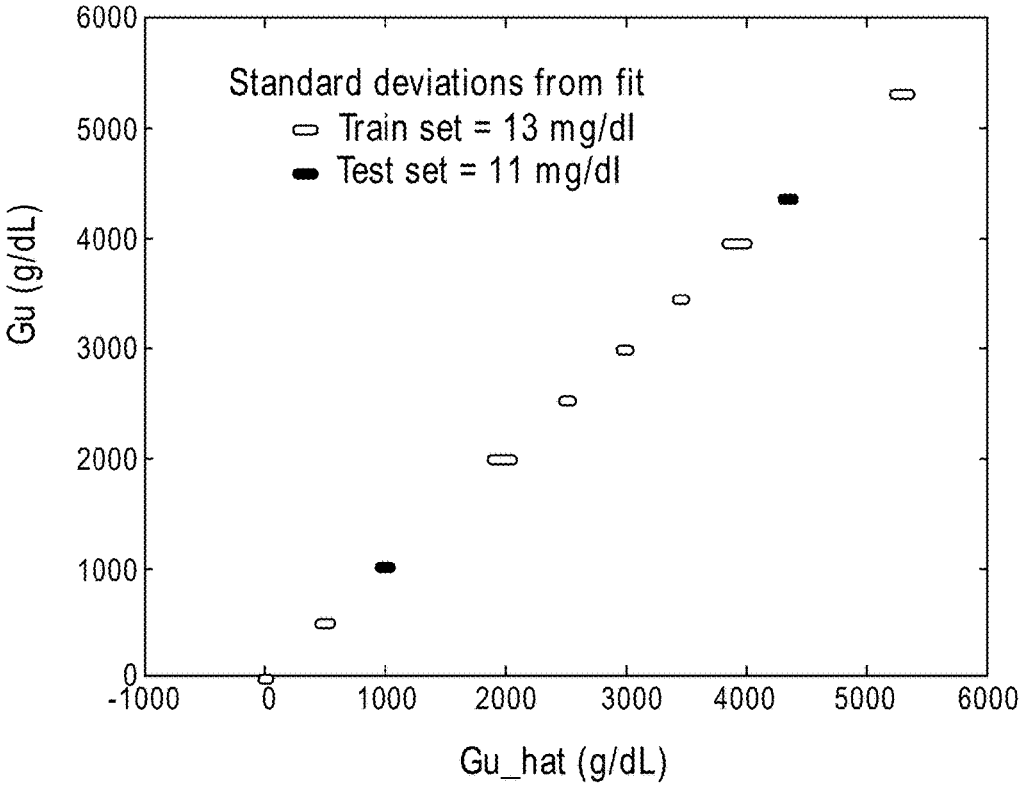

Standard deviations from fit
Train set = 13 mg/dl
Test set = 11 mg/dl
Gu (g/dL)
Gu_hat (g/dL)
6000
5000
4000
3000
2000
1000
0
-1000
0
1000
2000
3000
4000
5000
6000

FIG. 18

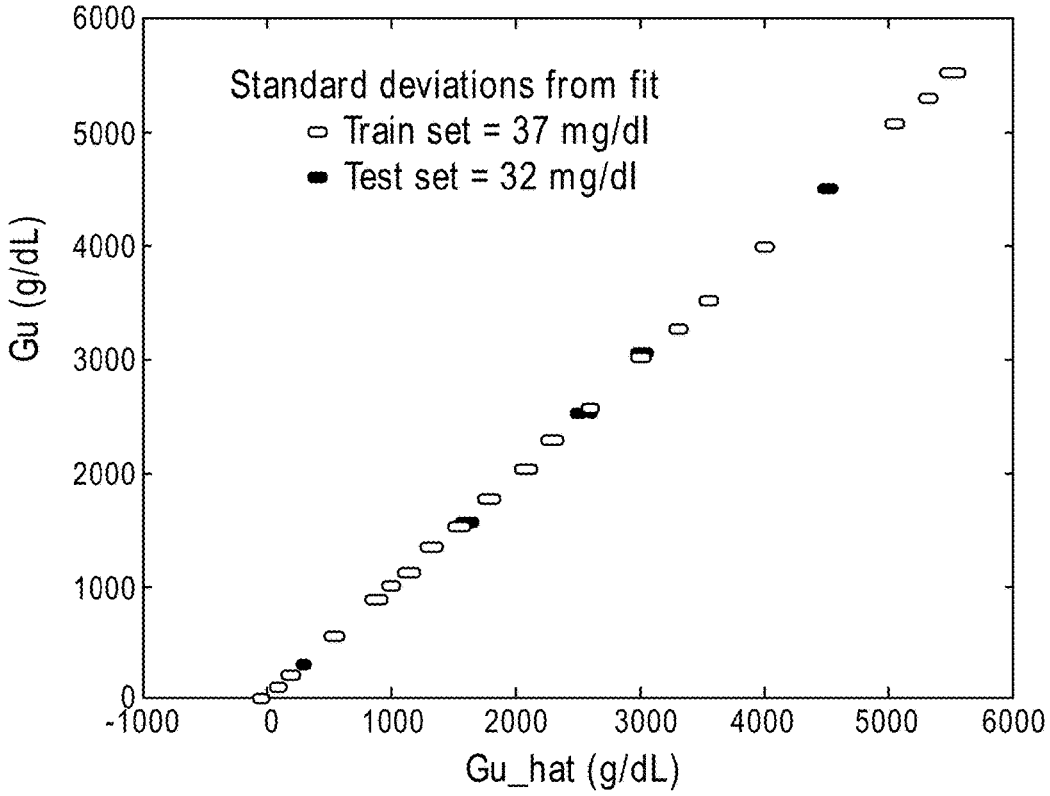

Standard deviations from fit
Train set = 37 mg/dl
Test set = 32 mg/dl
Gu (g/dL)
Gu_hat (g/dL)
6000
5000
4000
3000
2000
1000
0
-1000
0
1000
2000
3000
4000
5000
6000

FIG. 19

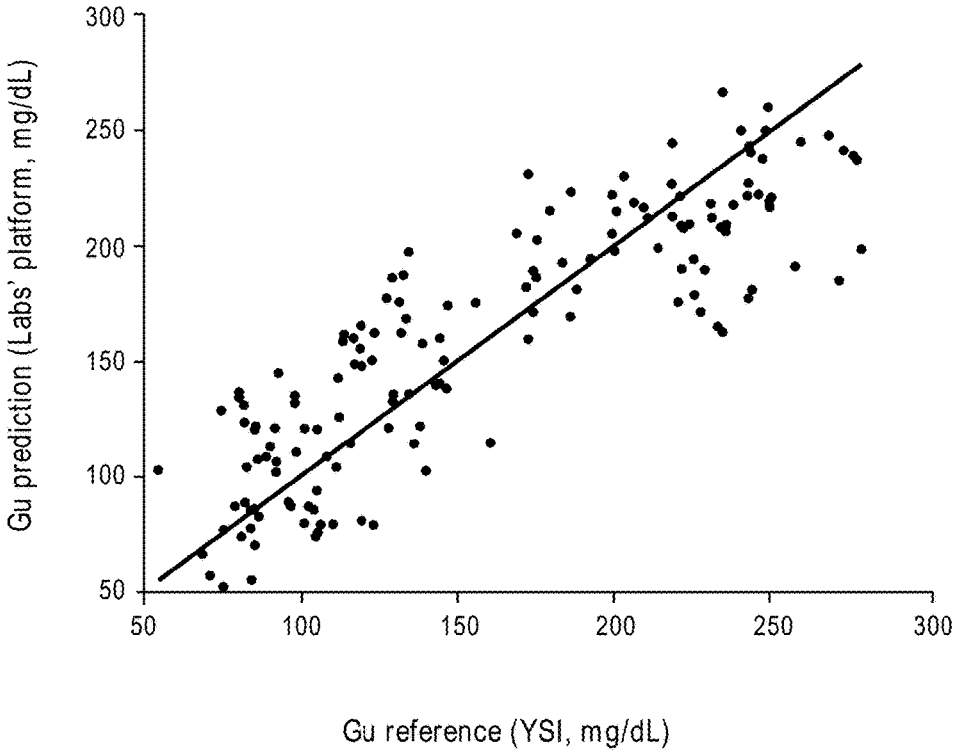

300
250
200
150
100
50
Gu prediction (Labs' platform, mg/dL)
50
100
150
200
250
300
Gu reference (YSI, mg/dL)

FIG. 20

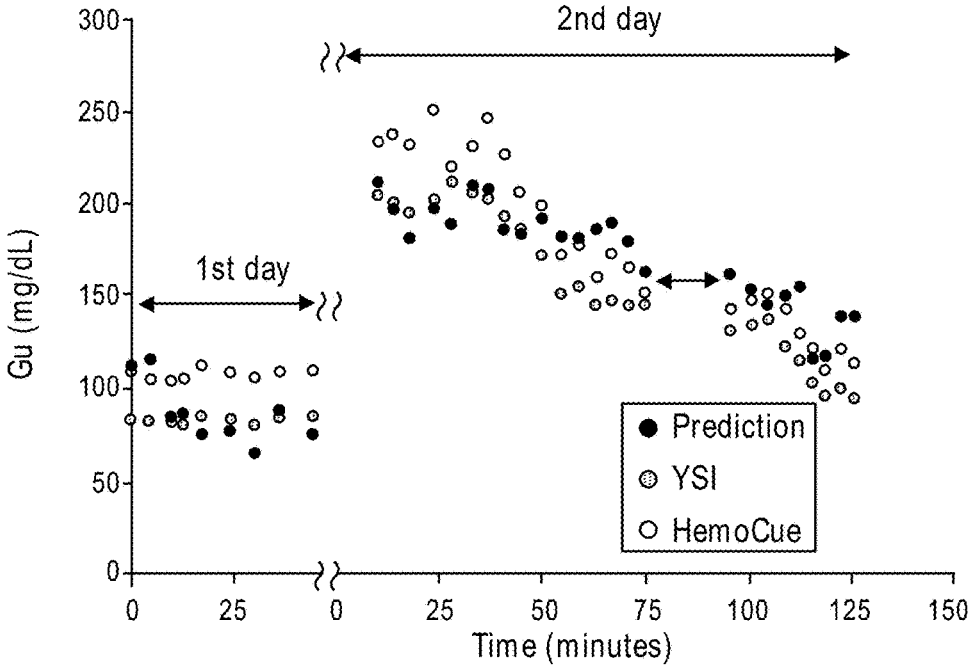

2nd day
1st day
Gu (mg/dL)
300
250
200
150
100
50
0
0
25
0
25
50
75
100
125
150
Time (minutes)
Prediction
YSI
HemoCue

FIG. 21

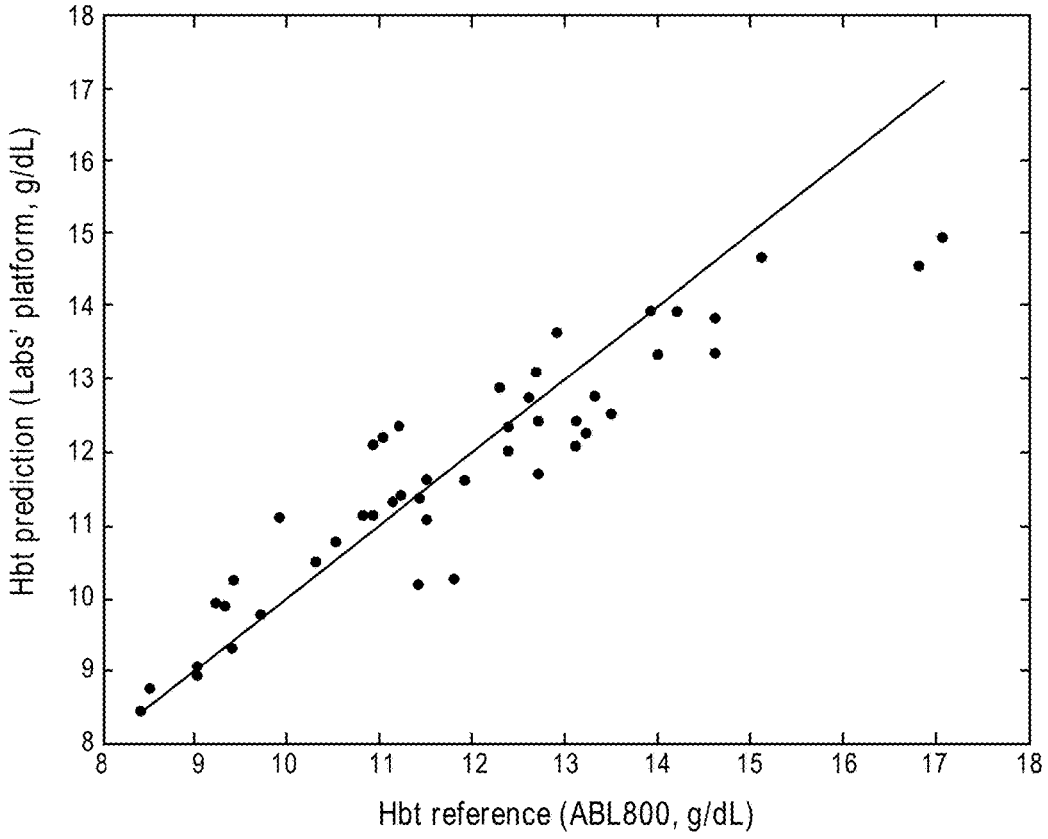

Hbt prediction (Labs' platform, g/dL)
Hbt reference (ABL800, g/dL)
8
9
10
11
12
13
14
15
16
17
18

FIG. 22

MULTI-STREAM DATA COLLECTION SYSTEM FOR NONINVASIVE MEASUREMENT OF BLOOD CONSTITUENTS

RELATED APPLICATIONS

This application is a continuation of U.S. patent application Ser. No. 14/981,290 filed Dec. 28, 2015, which is a continuation of U.S. patent application Ser. No. 12/829,352 filed Jul. 1, 2010, which is a continuation of U.S. patent application Ser. No. 12/534,827 filed Aug. 3, 2009, which claims the benefit of priority under 35 U.S.C. § 119(e) of the following U.S. Provisional Patent Application Nos. 61/086,060 filed Aug. 4, 2008, 61/086,108 filed Aug. 4, 2008, 61/086,063 filed Aug. 4, 2008, 61/086,057 filed Aug. 4, 2008, and 61/091,732 filed Aug. 25, 2008. U.S. patent application Ser. No. 12/829,352 is also a continuation-in-part of U.S. patent application Ser. No. 12/497,528 filed Jul. 2, 2009, which claims the benefit of priority under 35 U.S.C. § 119(e) of the following U.S. Provisional Patent Application Nos. 61/086,060 filed Aug. 4, 2008, 61/086,108 filed Aug. 4, 2008, 61/086,063 filed Aug. 4, 2008, 61/086,057 filed Aug. 4, 2008, 61/078,228 filed Jul. 3, 2008, 61/078,207 filed Jul. 3, 2008, and 61/091,732 filed Aug. 25, 2008. U.S. patent application Ser. No. 12/497,528 also claims the benefit of priority under 35 U.S.C. § 120 as a continuation-in-part of the following U.S. Design Patent application Nos. 29/323,409 filed Aug. 25, 2008 and Ser. No. 29/323,408 filed Aug. 25, 2008. U.S. patent application Ser. No. 12/829,352 is also a continuation-in-part of U.S. patent application Ser. No. 12/497,523 filed Jul. 2, 2009, which claims the benefit of priority under 35 U.S.C. § 119(e) of the following U.S. Provisional Patent Application Nos. 61/086,060 filed Aug. 4, 2008, 61/086,108 filed Aug. 4, 2008, 61/086,063 filed Aug. 4, 2008, 61/086,057 filed Aug. 4, 2008, 61/078,228 filed Jul. 3, 2008, 61/078,207 filed Jul. 3, 2008, and 61/091,732 filed Aug. 25, 2008. U.S. patent application Ser. No. 12/497,523 also claims the benefit of priority under 35 U.S.C. § 120 as a continuation-in-part of the following U.S. Design Patent application Nos. 29/323,409 filed Aug. 25, 2008 and Ser. No. 29/323,408 filed Aug. 25, 2008.

This application is related to the following U.S. Patent Applications:

App. No.	Filing Date	Title	Attorney Docket
12/497,528	Jul. 2, 2009	Noise Shielding for Noninvasive Device Contoured Protrusion for Improving	MASCER.006A
12/497,523	Jul. 2, 2009	Spectroscopic Measurement of Blood Constituents	MASCER.007A
12/497,506	Jul. 2, 2009	Heat Sink for Noninvasive Medical Sensor	MASCER.011A
12/534,812	Aug. 3, 2009	Multi-Stream Sensor Front Ends for Non-Invasive Measurement of Blood Constituents	MASCER.003A
12/534,823	Aug. 3, 2009	Multi-Stream Sensor for Non-Invasive Measurement of Blood Constituents	MASCER.004A
12/534,825	Aug. 3, 2009	Multi-Stream Emitter for Non-Invasive Measurement of Blood Constituents	CERCA.005A

The foregoing applications are hereby incorporated by reference in their entirety.

BACKGROUND

The standard of care in caregiver environments includes patient monitoring through spectroscopic analysis using, for example, a pulse oximeter. Devices capable of spectroscopic analysis generally include a light source(s) transmitting optical radiation into or reflecting off a measurement site, such as, body tissue carrying pulsing blood. After attenuation by tissue and fluids of the measurement site, a photodetection device(s) detects the attenuated light and outputs a detector signal(s) responsive to the detected attenuated light. A signal processing device(s) process the detector(s) signal (s) and outputs a measurement indicative of a blood constituent of interest, such as glucose, oxygen, met hemoglobin, total hemoglobin, other physiological parameters, or other data or combinations of data useful in determining a state or trend of wellness of a patient.

In noninvasive devices and methods, a sensor is often adapted to position a finger proximate the light source and light detector. For example, noninvasive sensors often include a clothespin-shaped housing that includes a contoured bed conforming generally to the shape of a finger.

SUMMARY

This disclosure describes embodiments of noninvasive methods, devices, and systems for measuring a blood constituent or analyte, such as oxygen, carbon monoxide, methemoglobin, total hemoglobin, glucose, proteins, glucose, lipids, a percentage thereof (e.g., saturation) or for measuring many other physiologically relevant patient characteristics. These characteristics can relate, for example, to pulse rate, hydration, trending information and analysis, and the like.

In an embodiment, the system includes a noninvasive sensor and a patient monitor communicating with the noninvasive sensor. The non-invasive sensor may include different architectures to implement some or all of the disclosed features. In addition, an artisan will recognize that the non-invasive sensor may include or may be coupled to other components, such as a network interface, and the like. Moreover, the patient monitor may include a display device, a network interface communicating with any one or combination of a computer network, a handheld computing device, a mobile phone, the Internet, or the like. In addition, embodiments may include multiple optical sources that emit light at a plurality of wavelengths and that are arranged from the perspective of the light detector(s) as a point source.

In an embodiment, a noninvasive device is capable of producing a signal responsive to light attenuated by tissue at a measurement site. The device may comprise an optical source and a plurality of photodetectors. The optical source is configured to emit optical radiation at least at wavelengths between about 1600 nm and about 1700 nm. The photode-

tectors are configured to detect the optical radiation from said optical source after attenuation by the tissue of the measurement site and each output a respective signal stream responsive to the detected optical radiation.

In an embodiment, a noninvasive, physiological sensor is capable of outputting a signal responsive to a blood analyte present in a monitored patient. The sensor may comprise a sensor housing, an optical source, and photodetectors. The optical source is positioned by the housing with respect to a tissue site of a patient when said housing is applied to the patient. The photodetectors are positioned by the housing with respect to said tissue site when the housing is applied to the patient with a variation in path length among at least some of the photodetectors from the optical source. The photodetectors are configured to detect a sequence of optical radiation from the optical source after attenuation by tissue of the tissue site. The photodetectors may be each configured to output a respective signal stream responsive to the detected sequence of optical radiation. An output signal responsive to one or more of the signal streams is then usable to determine the blood analyte based at least in part on the variation in path length

In an embodiment, a method of measuring an analyte based on multiple streams of optical radiation measured from a measurement site is provided. A sequence of optical radiation pulses is emitted to the measurement site. At a first location, a first stream of optical radiation is detected from the measurement site. At least at one additional location different from the first location, an additional stream of optical radiation is detected from the measurement site. An output measurement value indicative of the analyte is then determined based on the detected streams of optical radiation.

In various embodiments, the present disclosure relates to an interface for a noninvasive sensor that comprises a front-end adapted to receive an input signals from optical detectors and provide corresponding output signals. In an embodiment, the front-end is comprised of switched-capacitor circuits that are capable of handling multiple streams of signals from the optical detectors. In another embodiment, the front-end comprises transimpedance amplifiers that are capable of handling multiple streams of input signals. In addition, the transimpedance amplifiers may be configured based on the characteristics of the transimpedance amplifier itself, the characteristics of the photodiodes, and the number of photodiodes coupled to the transimpedance amplifier.

In disclosed embodiments, the front-ends are employed in noninvasive sensors to assist in measuring and detecting various analytes. The disclosed noninvasive sensor may also include, among other things, emitters and detectors positioned to produce multi-stream sensor information. An artisan will recognize that the noninvasive sensor may have different architectures and may include or be coupled to other components, such as a display device, a network interface, and the like. An artisan will also recognize that the front-ends may be employed in any type of noninvasive sensor.

In an embodiment, a front-end interface for a noninvasive, physiological sensor comprises: a set of inputs configured to receive signals from a plurality of detectors in the sensor; a set of transimpedance amplifiers configured to convert the signals from the plurality of detectors into an output signal having a stream for each of the plurality of detectors; and an output configured to provide the output signal.

In an embodiment, a front-end interface for a noninvasive, physiological sensor comprises: a set of inputs configured to receive signals from a plurality of detectors in the sensor; a set of switched capacitor circuits configured to convert the signals from the plurality of detectors into a digital output signal having a stream for each of the plurality of detectors; and an output configured to provide the digital output signal.

In an embodiment, a conversion processor for a physiological, noninvasive sensor comprises: a multi-stream input configured to receive signals from a plurality of detectors in the sensor, wherein the signals are responsive to optical radiation from a tissue site; a modulator that converts the multi-stream input into a digital bit-stream; and a signal processor that produces an output signal from the digital bit-stream.

In an embodiment, a front-end interface for a noninvasive, physiological sensor comprises: a set of inputs configured to receive signals from a plurality of detectors in the sensor; a set of respective transimpedance amplifiers for each detector configured to convert the signals from the plurality of detectors into an output signal having a stream for each of the plurality of detectors; and an output configured to provide the output signal.

In certain embodiments, a noninvasive sensor interfaces with tissue at a measurement site and deforms the tissue in a way that increases signal gain in certain desired wavelengths.

In some embodiments, a detector for the sensor may comprise a set of photodiodes that are arranged in a spatial configuration. This spatial configuration may allow, for example, signal analysis for measuring analytes like glucose. In various embodiments, the detectors can be arranged across multiple locations in a spatial configuration. The spatial configuration provides a geometry having a diversity of path lengths among the detectors. For example, the detector in the sensor may comprise multiple detectors that are arranged to have a sufficient difference in mean path length to allow for noise cancellation and noise reduction.

In an embodiment, a physiological, noninvasive detector is configured to detect optical radiation from a tissue site. The detector comprises a set of photodetectors and a conversion processor. The set of photodetectors each provide a signal stream indicating optical radiation from the tissue site. The set of photodetectors are arranged in a spatial configuration that provides a variation in path lengths between at least some of the photodetectors. The conversion processor that provides information indicating an analyte in the tissue site based on ratios of pairs of the signal streams.

The present disclosure, according to various embodiments, relates to noninvasive methods, devices, and systems for measuring a blood analyte, such as glucose. In the present disclosure, blood analytes are measured noninvasively based on multi-stream infrared and near-infrared spectroscopy. In some embodiments, an emitter may include one or more sources that are configured as a point optical source. In addition, the emitter may be operated in a manner that allows for the measurement of an analyte like glucose. In embodiments, the emitter may comprise a plurality of LEDs that emit a sequence of pulses of optical radiation across a spectrum of wavelengths. In addition, in order to achieve the desired SNR for detecting analytes like glucose, the emitter may be driven using a progression from low power to higher power. The emitter may also have its duty cycle modified to achieve a desired SNR.

In an embodiment, a multi-stream emitter for a noninvasive, physiological device configured to transmit optical radiation in a tissue site comprises: a set of optical sources arranged as a point optical source; and a driver configured to drive the at least one light emitting diode and at least one optical source to transmit near-infrared optical radiation at

sufficient power to measure an analyte in tissue that responds to near-infrared optical radiation.

In an embodiment, an emitter for a noninvasive, physiological device configured to transmit optical radiation in a tissue site comprises: a point optical source comprising an optical source configured to transmit infrared and near-infrared optical radiation to a tissue site; and a driver configured to drive the point optical source at a sufficient power and noise tolerance to effectively provide attenuated optical radiation from a tissue site that indicates an amount of glucose in the tissue site.

In an embodiment, a method of transmitting a stream of pulses of optical radiation in a tissue site is provided. At least one pulse of infrared optical radiation having a first pulse width is transmitted at a first power. At least one pulse of near-infrared optical radiation is transmitted at a power that is higher than the first power.

In an embodiment, a method of transmitting a stream of pulses of optical radiation in a tissue site is provided. At least one pulse of infrared optical radiation having a first pulse width is transmitted at a first power. At least one pulse of near-infrared optical radiation is then transmitted, at a second power that is higher than the first power.

For purposes of summarizing the disclosure, certain aspects, advantages and novel features of the inventions have been described herein. It is to be understood that not necessarily all such advantages can be achieved in accordance with any particular embodiment of the inventions disclosed herein. Thus, the inventions disclosed herein can be embodied or carried out in a manner that achieves or optimizes one advantage or group of advantages as taught herein without necessarily achieving other advantages as can be taught or suggested herein.

BRIEF DESCRIPTION OF THE DRAWINGS

Throughout the drawings, reference numbers can be reused to indicate correspondence between referenced elements. The drawings are provided to illustrate embodiments of the inventions described herein and not to limit the scope thereof.

FIG. **1** illustrates a block diagram of an example data collection system capable of noninvasively measuring one or more blood analytes in a monitored patient, according to an embodiment of the disclosure;

FIGS. **2**A-**2**D illustrate an exemplary handheld monitor and an exemplary noninvasive optical sensor of the patient monitoring system of FIG. **1**, according to embodiments of the disclosure;

FIGS. **3**A-**3**C illustrate side and perspective views of an exemplary noninvasive sensor housing including a finger bed protrusion and heat sink, according to an embodiment of the disclosure;

FIG. **3**D illustrates a side view of another example noninvasive sensor housing including a heat sink, according to an embodiment of the disclosure;

FIG. **3**E illustrates a perspective view of an example noninvasive sensor detector shell including example detectors, according to an embodiment of the disclosure;

FIG. **3**F illustrates a side view of an example noninvasive sensor housing including a finger bed protrusion and heat sink, according to an embodiment of the disclosure;

FIGS. **4**A through **4**C illustrate top elevation, side and top perspective views of an example protrusion, according to an embodiment of the disclosure;

FIG. **5** illustrates an example graph depicting possible effects of a protrusion on light transmittance, according to an embodiment of the disclosure;

FIGS. **6**A through **6**D illustrate perspective, front elevation, side and top views of another example protrusion, according to an embodiment of the disclosure;

FIG. **6**E illustrates an example sensor incorporating the protrusion of FIGS. **6**A through **6**D, according to an embodiment of the disclosure;

FIGS. **7**A through **7**B illustrate example arrangements of conductive glass that may be employed in the system of FIG. **1**, according to embodiments of the disclosure.

FIGS. **8**A through **8**D illustrate an example top elevation view, side views, and a bottom elevation view of the conductive glass that may be employed in the system of FIG. **1**, according to embodiments of the disclosure;

FIG. **9** shows example comparative results obtained by an embodiment of a sensor;

FIGS. **10**A and **10**B illustrate comparative noise floors of various embodiments of the present disclosure;

FIG. **11**A illustrates an exemplary emitter that may be employed in the sensor, according to an embodiment of the disclosure;

FIG. **11**B illustrates a configuration of emitting optical radiation into a measurement site for measuring blood constituents, according to an embodiment of the disclosure;

FIG. **11**C illustrates another exemplary emitter that may be employed in the sensor according to an embodiment of the disclosure;

FIG. **11**D illustrates another exemplary emitter that may be employed in the sensor according to an embodiment of the disclosure.

FIG. **12**A illustrates an example detector portion that may be employed in an embodiment of a sensor, according to an embodiment of the disclosure;

FIGS. **12**B through **12**D illustrate exemplary arrangements of detectors that may be employed in an embodiment of the sensor, according to some embodiments of the disclosure;

FIGS. **12**E through **12**H illustrate exemplary structures of photodiodes that may be employed in embodiments of the detectors, according to some embodiments of the disclosure;

FIG. **13** illustrates an example multi-stream operation of the system of FIG. **1**, according to an embodiment of the disclosure;

FIG. **14**A illustrates another example detector portion having a partially cylindrical protrusion that can be employed in an embodiment of a sensor, according to an embodiment of the disclosure;

FIG. **14**B depicts a front elevation view of the partially cylindrical protrusion of FIG. **14**A;

FIGS. **14**C through **14**E illustrate embodiments of a detector submount;

FIGS. **14**F through **14**H illustrate embodiment of portions of a detector shell;

FIG. **14**I illustrates a cutaway view of an embodiment of a sensor;

FIGS. **15**A through **15**F illustrate embodiments of sensors that include heat sink features;

FIGS. **15**G and **15**H illustrate embodiments of connector features that can be used with any of the sensors described herein;

FIG. **15**I illustrates an exemplary architecture for a transimpedance-based front-end that may be employed in any of the sensors described herein;

FIG. **15**J illustrates an exemplary noise model for configuring the transimpedance-based front-ends shown in FIG. **15**I;

FIG. **15**K shows different architectures and layouts for various embodiments of a sensor and its detectors;

FIG. **15**L illustrates an exemplary architecture for a switched-capacitor-based front-end that may be employed in any of the sensors described herein;

FIGS. **16**A and **16**B illustrate embodiments of disposable optical sensors;

FIG. **17** illustrates an exploded view of certain components of an example sensor; and

FIGS. **18** through **22** illustrate various results obtained by an exemplary sensor of the disclosure.

DETAILED DESCRIPTION

The present disclosure generally relates to non-invasive medical devices. In the present disclosure, a sensor can measure various blood constituents or analytes noninvasively using multi-stream spectroscopy. In an embodiment, the multi-stream spectroscopy can employ visible, infrared and near infrared wavelengths. As disclosed herein, the sensor is capable of noninvasively measuring blood analytes or percentages thereof (e.g., saturation) based on various combinations of features and components.

In various embodiments, the present disclosure relates to an interface for a noninvasive glucose sensor that comprises a front-end adapted to receive an input signals from optical detectors and provide corresponding output signals. The front-end may comprise, among other things, switched capacitor circuits or transimpedance amplifiers. In an embodiment, the front-end may comprise switched capacitor circuits that are configured to convert the output of sensor's detectors into a digital signal. In another embodiment, the front-end may comprise transimpedance amplifiers. These transimpedance amplifiers may be configured to match one or more photodiodes in a detector based on a noise model that accounts for characteristics, such as the impedance, of the transimpedance amplifier, characteristics of each photodiode, such as the impedance, and the number of photodiodes coupled to the transimpedance amplifier.

In the present disclosure, the front-ends are employed in a sensor that measures various blood analytes noninvasively using multi-stream spectroscopy. In an embodiment, the multi-stream spectroscopy can employ visible, infrared and near infrared wavelengths. As disclosed herein, the sensor is capable of noninvasively measuring blood analytes, such as glucose, total hemoglobin, methemoglobin, oxygen content, and the like, based on various combinations of features and components.

In an embodiment, a physiological sensor includes a detector housing that can be coupled to a measurement site, such as a patient's finger. The sensor housing can include a curved bed that can generally conform to the shape of the measurement site. In addition, the curved bed can include a protrusion shaped to increase an amount of light radiation from the measurement site. In an embodiment, the protrusion is used to thin out the measurement site. This allows the light radiation to pass through less tissue, and accordingly is attenuated less. In an embodiment, the protrusion can be used to increase the area from which attenuated light can be measured. In an embodiment, this is done through the use of a lens which collects attenuated light exiting the measurement site and focuses onto one or more detectors. The protrusion can advantageously include plastic, including a hard opaque plastic, such as a black or other colored plastic, helpful in reducing light noise. In an embodiment, such light noise includes light that would otherwise be detected at a photodetector that has not been attenuated by tissue of the measurement site of a patient sufficient to cause the light to adequately included information indicative of one or more physiological parameters of the patient. Such light noise includes light piping.

In an embodiment, the protrusion can be formed from the curved bed, or can be a separate component that is positionable with respect to the bed. In an embodiment, a lens made from any appropriate material is used as the protrusion. The protrusion can be convex in shape. The protrusion can also be sized and shaped to conform the measurement site into a flat or relatively flat surface. The protrusion can also be sized to conform the measurement site into a rounded surface, such as, for example, a concave or convex surface. The protrusion can include a cylindrical or partially cylindrical shape. The protrusion can be sized or shaped differently for different types of patients, such as an adult, child, or infant. The protrusion can also be sized or shaped differently for different measurement sites, including, for example, a finger, toe, hand, foot, ear, forehead, or the like. The protrusion can thus be helpful in any type of noninvasive sensor. The external surface of the protrusion can include one or more openings or windows. The openings can be made from glass to allow attenuated light from a measurement site, such as a finger, to pass through to one or more detectors. Alternatively, some of all of the protrusion can be a lens, such as a partially cylindrical lens.

The sensor can also include a shielding, such as a metal enclosure as described below or embedded within the protrusion to reduce noise. The shielding can be constructed from a conductive material, such as copper, in the form of a metal cage or enclosure, such as a box. The shielding can include a second set of one or more openings or windows. The second set of openings can be made from glass and allow light that has passed through the first set of windows of the external surface of the protrusion to pass through to one or more detectors that can be enclosed, for example, as described below.

In various embodiments, the shielding can include any substantially transparent, conductive material placed in the optical path between an emitter and a detector. The shielding can be constructed from a transparent material, such as glass, plastic, and the like. The shielding can have an electrically conductive material or coating that is at least partially transparent. The electrically conductive coating can be located on one or both sides of the shielding, or within the body of the shielding. In addition, the electrically conductive coating can be uniformly spread over the shielding or may be patterned. Furthermore, the coating can have a uniform or varying thickness to increase or optimize its shielding effect. The shielding can be helpful in virtually any type of noninvasive sensor that employs spectroscopy.

In an embodiment, the sensor can also include a heat sink. In an embodiment, the heat sink can include a shape that is functional in its ability to dissipate excess heat and aesthetically pleasing to the wearer. For example, the heat sink can be configured in a shape that maximizes surface area to allow for greater dissipation of heat. In an embodiment, the heat sink includes a metalicized plastic, such as plastic including carbon and aluminum to allow for improved thermal conductivity and diffusivity. In an embodiment, the heat sink can advantageously be inexpensively molded into desired shapes and configurations for aesthetic and functional purposes. For example, the shape of the heat sink can

be a generally curved surface and include one or more fins, undulations, grooves or channels, or combs.

The sensor can include photocommunicative components, such as an emitter, a detector, and other components. The emitter can include a plurality of sets of optical sources that, in an embodiment, are arranged together as a point source. The various optical sources can emit a sequence of optical radiation pulses at different wavelengths towards a measurement site, such as a patient's finger. Detectors can then detect optical radiation from the measurement site. The optical sources and optical radiation detectors can operate at any appropriate wavelength, including, as discussed herein, infrared, near infrared, visible light, and ultraviolet. In addition, the optical sources and optical radiation detectors can operate at any appropriate wavelength, and such modifications to the embodiments desirable to operate at any such wavelength will be apparent to those skilled in the art.

In certain embodiments, multiple detectors are employed and arranged in a spatial geometry. This spatial geometry provides a diversity of path lengths among at least some of the detectors and allows for multiple bulk and pulsatile measurements that are robust. Each of the detectors can provide a respective output stream based on the detected optical radiation, or a sum of output streams can be provided from multiple detectors. In some embodiments, the sensor can also include other components, such as one or more heat sinks and one or more thermistors.

The spatial configuration of the detectors provides a geometry having a diversity of path lengths among the detectors. For example, a detector in the sensor may comprise multiple detectors that are arranged to have a sufficient difference in mean path length to allow for noise cancellation and noise reduction. In addition, walls may be used to separate individual photodetectors and prevent mixing of detected optical radiation between the different locations on the measurement site. A window may also be employed to facilitate the passing of optical radiation at various wavelengths for measuring glucose in the tissue.

In the present disclosure, a sensor may measure various blood constituents or analytes noninvasively using spectroscopy and a recipe of various features. As disclosed herein, the sensor is capable of non-invasively measuring blood analytes, such as, glucose, total hemoglobin, methemoglobin, oxygen content, and the like. In an embodiment, the spectroscopy used in the sensor can employ visible, infrared and near infrared wavelengths. The sensor may comprise an emitter, a detector, and other components. In some embodiments, the sensor may also comprise other components, such as one or more heat sinks and one or more thermistors.

In various embodiments, the sensor may also be coupled to one or more companion devices that process and/or display the sensor's output. The companion devices may comprise various components, such as a sensor front-end, a signal processor, a display, a network interface, a storage device or memory, etc.

A sensor can include photocommunicative components, such as an emitter, a detector, and other components. The emitter is configured as a point optical source that comprises a plurality of LEDs that emit a sequence of pulses of optical radiation across a spectrum of wavelengths. In some embodiments, the plurality of sets of optical sources may each comprise at least one top-emitting LED and at least one super luminescent LED. In some embodiments, the emitter comprises optical sources that transmit optical radiation in the infrared or near-infrared wavelengths suitable for detecting blood analytes like glucose. In order to achieve the desired SNR for detecting analytes like glucose, the emitter may be driven using a progression from low power to higher power. In addition, the emitter may have its duty cycle modified to achieve a desired SNR.

The emitter may be constructed of materials, such as aluminum nitride and may include a heat sink to assist in heat dissipation. A thermistor may also be employed to account for heating effects on the LEDs. The emitter may further comprise a glass window and a nitrogen environment to improve transmission from the sources and prevent oxidative effects.

The sensor can be coupled to one or more monitors that process and/or display the sensor's output. The monitors can include various components, such as a sensor front end, a signal processor, a display, etc.

The sensor can be integrated with a monitor, for example, into a handheld unit including the sensor, a display and user controls. In other embodiments, the sensor can communicate with one or more processing devices. The communication can be via wire(s), cable(s), flex circuit(s), wireless technologies, or other suitable analog or digital communication methodologies and devices to perform those methodologies. Many of the foregoing arrangements allow the sensor to be attached to the measurement site while the device is attached elsewhere on a patient, such as the patient's arm, or placed at a location near the patient, such as a bed, shelf or table. The sensor or monitor can also provide outputs to a storage device or network interface.

Reference will now be made to the Figures to discuss embodiments of the present disclosure.

FIG. **1** illustrates an example of a data collection system **100**. In certain embodiments, the data collection system **100** noninvasively measure a blood analyte, such as oxygen, carbon monoxide, methemoglobin, total hemoglobin, glucose, proteins, glucose, lipids, a percentage thereof (e.g., saturation) or for measuring many other physiologically relevant patient characteristics. The system **100** can also measure additional blood analytes and/or other physiological parameters useful in determining a state or trend of wellness of a patient.

The data collection system **100** can be capable of measuring optical radiation from the measurement site. For example, in some embodiments, the data collection system **100** can employ photodiodes defined in terms of area. In an embodiment, the area is from about 1 mm^2-5 mm^2 (or higher) that are capable of detecting about 100 nanoamps (nA) or less of current resulting from measured light at full scale. In addition to having its ordinary meaning, the phrase "at full scale" can mean light saturation of a photodiode amplifier (not shown). Of course, as would be understood by a person of skill in the art from the present disclosure, various other sizes and types of photodiodes can be used with the embodiments of the present disclosure.

The data collection system **100** can measure a range of approximately about 2 nA to about 100 nA full scale. The data collection system **100** can also include sensor front-ends that are capable of processing and amplifying current from the detector(s) at signal-to-noise ratios (SNRs) of about 100 decibels (dB) or more, such as about 120 dB in order to measure various desired analytes. The data collection system **100** can operate with a lower SNR if less accuracy is desired for an analyte like glucose.

The data collection system **100** can measure analyte concentrations, including glucose, at least in part by detecting light attenuated by a measurement site **102**. The measurement site **102** can be any location on a patient's body, such as a finger, foot, ear lobe, or the like. For convenience, this disclosure is described primarily in the context of a

finger measurement site **102**. However, the features of the embodiments disclosed herein can be used with other measurement sites **102**.

In the depicted embodiment, the system **100** includes an optional tissue thickness adjuster or tissue shaper **105**, which can include one or more protrusions, bumps, lenses, or other suitable tissue-shaping mechanisms. In certain embodiments, the tissue shaper **105** is a flat or substantially flat surface that can be positioned proximate the measurement site **102** and that can apply sufficient pressure to cause the tissue of the measurement site **102** to be flat or substantially flat. In other embodiments, the tissue shaper **105** is a convex or substantially convex surface with respect to the measurement site **102**. Many other configurations of the tissue shaper **105** are possible. Advantageously, in certain embodiments, the tissue shaper **105** reduces thickness of the measurement site **102** while preventing or reducing occlusion at the measurement site **102**. Reducing thickness of the site can advantageously reduce the amount of attenuation of the light because there is less tissue through which the light must travel. Shaping the tissue in to a convex (or alternatively concave) surface can also provide more surface area from which light can be detected.

The embodiment of the data collection system **100** shown also includes an optional noise shield **103**. In an embodiment, the noise shield **103** can be advantageously adapted to reduce electromagnetic noise while increasing the transmittance of light from the measurement site **102** to one or more detectors **106** (described below). For example, the noise shield **103** can advantageously include a conductive coated glass or metal grid electrically communicating with one or more other shields of the sensor **101** or electrically grounded. In an embodiment where the noise shield **103** includes conductive coated glass, the coating can advantageously include indium tin oxide. In an embodiment, the indium tin oxide includes a surface resistivity ranging from approximately 30 ohms per square inch to about 500 ohms per square inch. In an embodiment, the resistivity is approximately 30, 200, or 500 ohms per square inch. As would be understood by a person of skill in the art from the present disclosure, other resistivities can also be used which are less than about 30 ohms or more than about 500 ohms. Other conductive materials transparent or substantially transparent to light can be used instead.

In some embodiments, the measurement site **102** is located somewhere along a non-dominant arm or a non-dominant hand, e.g., a right-handed person's left arm or left hand. In some patients, the non-dominant arm or hand can have less musculature and higher fat content, which can result in less water content in that tissue of the patient. Tissue having less water content can provide less interference with the particular wavelengths that are absorbed in a useful manner by blood analytes like glucose. Accordingly, in some embodiments, the data collection system **100** can be used on a person's non-dominant hand or arm.

The data collection system **100** can include a sensor **101** (or multiple sensors) that is coupled to a processing device or physiological monitor **109**. In an embodiment, the sensor **101** and the monitor **109** are integrated together into a single unit. In another embodiment, the sensor **101** and the monitor **109** are separate from each other and communicate one with another in any suitable manner, such as via a wired or wireless connection. The sensor **101** and monitor **109** can be attachable and detachable from each other for the convenience of the user or caregiver, for ease of storage, sterility issues, or the like. The sensor **101** and the monitor **109** will now be further described.

In the depicted embodiment shown in FIG. **1**, the sensor **101** includes an emitter **104**, a tissue shaper **105**, a set of detectors **106**, and a front-end interface **108**. The emitter **104** can serve as the source of optical radiation transmitted towards measurement site **102**. As will be described in further detail below, the emitter **104** can include one or more sources of optical radiation, such as LEDs, laser diodes, incandescent bulbs with appropriate frequency-selective filters, combinations of the same, or the like. In an embodiment, the emitter **104** includes sets of optical sources that are capable of emitting visible and near-infrared optical radiation.

In some embodiments, the emitter **104** is used as a point optical source, and thus, the one or more optical sources of the emitter **104** can be located within a close distance to each other, such as within about a 2 mm to about 4 mm. The emitters **104** can be arranged in an array, such as is described in U.S. Publication No. 2006/0211924, filed Sep. 21, 2006, titled "Multiple Wavelength Sensor Emitters," the disclosure of which is hereby incorporated by reference in its entirety. In particular, the emitters **104** can be arranged at least in part as described in paragraphs [0061] through [0068] of the aforementioned publication, which paragraphs are hereby incorporated specifically by reference. Other relative spatial relationships can be used to arrange the emitters **104**.

For analytes like glucose, currently available non-invasive techniques often attempt to employ light near the water absorbance minima at or about 1600 nm. Typically, these devices and methods employ a single wavelength or single band of wavelengths at or about 1600 nm. However, to date, these techniques have been unable to adequately consistently measure analytes like glucose based on spectroscopy.

In contrast, the emitter **104** of the data collection system **100** can emit, in certain embodiments, combinations of optical radiation in various bands of interest. For example, in some embodiments, for analytes like glucose, the emitter **104** can emit optical radiation at three (3) or more wavelengths between about 1600 nm to about 1700 nm. In particular, the emitter **104** can emit optical radiation at or about 1610 nm, about 1640 nm, and about 1665 nm. In some circumstances, the use of three wavelengths within about 1600 nm to about 1700 nm enable sufficient SNRs of about 100 dB, which can result in a measurement accuracy of about 20 mg/dL or better for analytes like glucose.

In other embodiments, the emitter **104** can use two (2) wavelengths within about 1600 nm to about 1700 nm to advantageously enable SNRs of about 85 dB, which can result in a measurement accuracy of about 25-30 mg/dL or better for analytes like glucose. Furthermore, in some embodiments, the emitter **104** can emit light at wavelengths above about 1670 nm. Measurements at these wavelengths can be advantageously used to compensate or confirm the contribution of protein, water, and other non-hemoglobin species exhibited in measurements for analytes like glucose conducted between about 1600 nm and about 1700 nm. Of course, other wavelengths and combinations of wavelengths can be used to measure analytes and/or to distinguish other types of tissue, fluids, tissue properties, fluid properties, combinations of the same or the like.

For example, the emitter **104** can emit optical radiation across other spectra for other analytes. In particular, the emitter **104** can employ light wavelengths to measure various blood analytes or percentages (e.g., saturation) thereof. For example, in one embodiment, the emitter **104** can emit optical radiation in the form of pulses at wavelengths about 905 nm, about 1050 nm, about 1200 nm, about 1300 nm, about 1330 nm, about 1610 nm, about 1640 nm, and about

1665 nm. In another embodiment, the emitter **104** can emit optical radiation ranging from about 860 nm to about 950 nm, about 950 nm to about 1100 nm, about 1100 nm to about 1270 nm, about 1250 nm to about 1350 nm, about 1300 nm to about 1360 nm, and about 1590 nm to about 1700 nm. Of course, the emitter **104** can transmit any of a variety of wavelengths of visible or near-infrared optical radiation.

Due to the different responses of analytes to the different wavelengths, certain embodiments of the data collection system **100** can advantageously use the measurements at these different wavelengths to improve the accuracy of measurements. For example, the measurements of water from visible and infrared light can be used to compensate for water absorbance that is exhibited in the near-infrared wavelengths.

As briefly described above, the emitter **104** can include sets of light-emitting diodes (LEDs) as its optical source. The emitter **104** can use one or more top-emitting LEDs. In particular, in some embodiments, the emitter **104** can include top-emitting LEDs emitting light at about 850 nm to 1350 nm.

The emitter **104** can also use super luminescent LEDs (SLEDs) or side-emitting LEDs. In some embodiments, the emitter **104** can employ SLEDs or side-emitting LEDs to emit optical radiation at about 1600 nm to about 1800 nm. Emitter **104** can use SLEDs or side-emitting LEDs to transmit near infrared optical radiation because these types of sources can transmit at high power or relatively high power, e.g., about 40 mW to about 100 mW. This higher power capability can be useful to compensate or overcome the greater attenuation of these wavelengths of light in tissue and water. For example, the higher power emission can effectively compensate and/or normalize the absorption signal for light in the mentioned wavelengths to be similar in amplitude and/or effect as other wavelengths that can be detected by one or more photodetectors after absorption. However, the embodiments of the present disclosure do not necessarily require the use of high power optical sources. For example, some embodiments may be configured to measure analytes, such as total hemoglobin (tHb), oxygen saturation (SpO_2), carboxyhemoglobin, methemoglobin, etc., without the use of high power optical sources like side emitting LEDs. Instead, such embodiments may employ other types of optical sources, such as top emitting LEDs. Alternatively, the emitter **104** can use other types of sources of optical radiation, such as a laser diode, to emit near-infrared light into the measurement site **102**.

In addition, in some embodiments, in order to assist in achieving a comparative balance of desired power output between the LEDs, some of the LEDs in the emitter **104** can have a filter or covering that reduces and/or cleans the optical radiation from particular LEDs or groups of LEDs. For example, since some wavelengths of light can penetrate through tissue relatively well, LEDs, such as some or all of the top-emitting LEDs can use a filter or covering, such as a cap or painted dye. This can be useful in allowing the emitter **104** to use LEDs with a higher output and/or to equalize intensity of LEDs.

The data collection system **100** also includes a driver **111** that drives the emitter **104**. The driver **111** can be a circuit or the like that is controlled by the monitor **109**. For example, the driver **111** can provide pulses of current to the emitter **104**. In an embodiment, the driver **111** drives the emitter **104** in a progressive fashion, such as in an alternating manner. The driver **111** can drive the emitter **104** with a series of pulses of about 1 milliwatt (mW) for some wavelengths that can penetrate tissue relatively well and from about 40 mW to about 100 mW for other wavelengths that tend to be significantly absorbed in tissue. A wide variety of other driving powers and driving methodologies can be used in various embodiments.

The driver **111** can be synchronized with other parts of the sensor **101** and can minimize or reduce jitter in the timing of pulses of optical radiation emitted from the emitter **104**. In some embodiments, the driver **111** is capable of driving the emitter **104** to emit optical radiation in a pattern that varies by less than about 10 parts-per-million.

The detectors **106** capture and measure light from the measurement site **102**. For example, the detectors **106** can capture and measure light transmitted from the emitter **104** that has been attenuated or reflected from the tissue in the measurement site **102**. The detectors **106** can output a detector signal **107** responsive to the light captured or measured. The detectors **106** can be implemented using one or more photodiodes, phototransistors, or the like.

In addition, the detectors **106** can be arranged with a spatial configuration to provide a variation of path lengths among at least some of the detectors **106**. That is, some of the detectors **106** can have the substantially, or from the perspective of the processing algorithm, effectively, the same path length from the emitter **104**. However, according to an embodiment, at least some of the detectors **106** can have a different path length from the emitter **104** relative to other of the detectors **106**. Variations in path lengths can be helpful in allowing the use of a bulk signal stream from the detectors **106**. In some embodiments, the detectors **106** may employ a linear spacing, a logarithmic spacing, or a two or three dimensional matrix of spacing, or any other spacing scheme in order to provide an appropriate variation in path lengths.

The front end interface **108** provides an interface that adapts the output of the detectors **106**, which is responsive to desired physiological parameters. For example, the front end interface **108** can adapt a signal **107** received from one or more of the detectors **106** into a form that can be processed by the monitor **109**, for example, by a signal processor **110** in the monitor **109**. The front end interface **108** can have its components assembled in the sensor **101**, in the monitor **109**, in connecting cabling (if used), combinations of the same, or the like. The location of the front end interface **108** can be chosen based on various factors including space desired for components, desired noise reductions or limits, desired heat reductions or limits, and the like.

The front end interface **108** can be coupled to the detectors **106** and to the signal processor **110** using a bus, wire, electrical or optical cable, flex circuit, or some other form of signal connection. The front end interface **108** can also be at least partially integrated with various components, such as the detectors **106**. For example, the front end interface **108** can include one or more integrated circuits that are on the same circuit board as the detectors **106**. Other configurations can also be used.

The front end interface **108** can be implemented using one or more amplifiers, such as transimpedance amplifiers, that are coupled to one or more analog to digital converters (ADCs) (which can be in the monitor **109**), such as a sigma-delta ADC. A transimpedance-based front end interface **108** can employ single-ended circuitry, differential circuitry, and/or a hybrid configuration. A transimpedance-based front end interface **108** can be useful for its sampling rate capability and freedom in modulation/demodulation algorithms. For example, this type of front end interface **108**

can advantageously facilitate the sampling of the ADCs being synchronized with the pulses emitted from the emitter **104**.

The ADC or ADCs can provide one or more outputs into multiple channels of digital information for processing by the signal processor **110** of the monitor **109**. Each channel can correspond to a signal output from a detector **106**.

In some embodiments, a programmable gain amplifier (PGA) can be used in combination with a transimpedance-based front end interface **108**. For example, the output of a transimpedance-based front end interface **108** can be output to a PGA that is coupled with an ADC in the monitor **109**. A PGA can be useful in order to provide another level of amplification and control of the stream of signals from the detectors **106**. Alternatively, the PGA and ADC components can be integrated with the transimpedance-based front end interface **108** in the sensor **101**.

In another embodiment, the front end interface **108** can be implemented using switched-capacitor circuits. A switched-capacitor-based front end interface **108** can be useful for, in certain embodiments, its resistor-free design and analog averaging properties. In addition, a switched-capacitor-based front end interface **108** can be useful because it can provide a digital signal to the signal processor **110** in the monitor **109**.

As shown in FIG. **1**, the monitor **109** can include the signal processor **110** and a user interface, such as a display **112**. The monitor **109** can also include optional outputs alone or in combination with the display **112**, such as a storage device **114** and a network interface **116**. In an embodiment, the signal processor **110** includes processing logic that determines measurements for desired analytes, such as glucose, based on the signals received from the detectors **106**. The signal processor **110** can be implemented using one or more microprocessors or subprocessors (e.g., cores), digital signal processors, application specific integrated circuits (ASICs), field programmable gate arrays (FPGAs), combinations of the same, and the like.

The signal processor **110** can provide various signals that control the operation of the sensor **101**. For example, the signal processor **110** can provide an emitter control signal to the driver **111**. This control signal can be useful in order to synchronize, minimize, or reduce jitter in the timing of pulses emitted from the emitter **104**. Accordingly, this control signal can be useful in order to cause optical radiation pulses emitted from the emitter **104** to follow a precise timing and consistent pattern. For example, when a transimpedance-based front end interface **108** is used, the control signal from the signal processor **110** can provide synchronization with the ADC in order to avoid aliasing, cross-talk, and the like. As also shown, an optional memory **113** can be included in the front-end interface **108** and/or in the signal processor **110**. This memory **113** can serve as a buffer or storage location for the front-end interface **108** and/or the signal processor **110**, among other uses.

The user interface **112** can provide an output, e.g., on a display, for presentation to a user of the data collection system **100**. The user interface **112** can be implemented as a touch-screen display, an LCD display, an organic LED display, or the like. In addition, the user interface **112** can be manipulated to allow for measurement on the non-dominant side of patient. For example, the user interface **112** can include a flip screen, a screen that can be moved from one side to another on the monitor **109**, or can include an ability to reorient its display indicia responsive to user input or device orientation. In alternative embodiments, the data collection system **100** can be provided without a user interface **112** and can simply provide an output signal to a separate display or system.

A storage device **114** and a network interface **116** represent other optional output connections that can be included in the monitor **109**. The storage device **114** can include any computer-readable medium, such as a memory device, hard disk storage, EEPROM, flash drive, or the like. The various software and/or firmware applications can be stored in the storage device **114**, which can be executed by the signal processor **110** or another processor of the monitor **109**. The network interface **116** can be a serial bus port (RS-232/RS-485), a Universal Serial Bus (USB) port, an Ethernet port, a wireless interface (e.g., WiFi such as any 802.1x interface, including an internal wireless card), or other suitable communication device(s) that allows the monitor **109** to communicate and share data with other devices. The monitor **109** can also include various other components not shown, such as a microprocessor, graphics processor, or controller to output the user interface **112**, to control data communications, to compute data trending, or to perform other operations.

Although not shown in the depicted embodiment, the data collection system **100** can include various other components or can be configured in different ways. For example, the sensor **101** can have both the emitter **104** and detectors **106** on the same side of the measurement site **102** and use reflectance to measure analytes. The data collection system **100** can also include a sensor that measures the power of light emitted from the emitter **104**.

FIGS. **2**A through **2**D illustrate example monitoring devices **200** in which the data collection system **100** can be housed. Advantageously, in certain embodiments, some or all of the example monitoring devices **200** shown can have a shape and size that allows a user to operate it with a single hand or attach it, for example, to a patient's body or limb. Although several examples are shown, many other monitoring device configurations can be used to house the data collection system **100**. In addition, certain of the features of the monitoring devices **200** shown in FIGS. **2**A through **2**D can be combined with features of the other monitoring devices **200** shown.

Referring specifically to FIG. **2**A, an example monitoring device **200**A is shown, in which a sensor **201***a* and a monitor **209***a* are integrated into a single unit. The monitoring device **200**A shown is a handheld or portable device that can measure glucose and other analytes in a patient's finger. The sensor **201***a* includes an emitter shell **204***a* and a detector shell **206***a*. The depicted embodiment of the monitoring device **200**A also includes various control buttons **208***a* and a display **210***a*.

The sensor **201***a* can be constructed of white material used for reflective purposes (such as white silicone or plastic), which can increase the usable signal at the detector **106** by forcing light back into the sensor **201***a*. Pads in the emitter shell **204***a* and the detector shell **206***a* can contain separated windows to prevent or reduce mixing of light signals, for example, from distinct quadrants on a patient's finger. In addition, these pads can be made of a relatively soft material, such as a gel or foam, in order to conform to the shape, for example, of a patient's finger. The emitter shell **204***a* and the detector shell **206***a* can also include absorbing black or grey material portions to prevent or reduce ambient light from entering into the sensor **201***a*.

In some embodiments, some or all portions of the emitter shell **204***a* and/or detector shell **206***a* can be detachable and/or disposable. For example, some or all portions of the

shells **204***a* and **206***a* can be removable pieces. The removability of the shells **204***a* and **206***a* can be useful for sanitary purposes or for sizing the sensor **201***a* to different patients. The monitor **209***a* can include a fitting, slot, magnet, or other connecting mechanism to allow the sensor **201***c* to be removably attached to the monitor **209***a*.

The monitoring device **200***a* also includes optional control buttons **208***a* and a display **210***a* that can allow the user to control the operation of the device. For example, a user can operate the control buttons **208***a* to view one or more measurements of various analytes, such as glucose. In addition, the user can operate the control buttons **208***a* to view other forms of information, such as graphs, histograms, measurement data, trend measurement data, parameter combination views, wellness indications, and the like. Many parameters, trends, alarms and parameter displays could be output to the display **210***a*, such as those that are commercially available through a wide variety of noninvasive monitoring devices from Masimo® Corporation of Irvine, Calif.

Furthermore, the controls **208***a* and/or display **210***a* can provide functionality for the user to manipulate settings of the monitoring device **200***a*, such as alarm settings, emitter settings, detector settings, and the like. The monitoring device **200***a* can employ any of a variety of user interface designs, such as frames, menus, touch-screens, and any type of button.

FIG. **2**B illustrates another example of a monitoring device **200**B. In the depicted embodiment, the monitoring device **200**B includes a finger clip sensor **201***b* connected to a monitor **209***b* via a cable **212**. In the embodiment shown, the monitor **209***b* includes a display **210***b*, control buttons **208***b* and a power button. Moreover, the monitor **209***b* can advantageously include electronic processing, signal processing, and data storage devices capable of receiving signal data from said sensor **201***b*, processing the signal data to determine one or more output measurement values indicative of one or more physiological parameters of a monitored patient, and displaying the measurement values, trends of the measurement values, combinations of measurement values, and the like.

The cable **212** connecting the sensor **201***b* and the monitor **209***b* can be implemented using one or more wires, optical fiber, flex circuits, or the like. In some embodiments, the cable **212** can employ twisted pairs of conductors in order to minimize or reduce cross-talk of data transmitted from the sensor **201***b* to the monitor **209***b*. Various lengths of the cable **212** can be employed to allow for separation between the sensor **201***b* and the monitor **209***b*. The cable **212** can be fitted with a connector (male or female) on either end of the cable **212** so that the sensor **201***b* and the monitor **209***b* can be connected and disconnected from each other. Alternatively, the sensor **201***b* and the monitor **209***b* can be coupled together via a wireless communication link, such as an infrared link, radio frequency channel, or any other wireless communication protocol and channel.

The monitor **209***b* can be attached to the patient. For example, the monitor **209***b* can include a belt clip or straps (see, e.g., FIG. **2**C) that facilitate attachment to a patient's belt, arm, leg, or the like. The monitor **209***b* can also include a fitting, slot, magnet, LEMO snap-click connector, or other connecting mechanism to allow the cable **212** and sensor **201***b* to be attached to the monitor **209**B.

The monitor **209***b* can also include other components, such as a speaker, power button, removable storage or memory (e.g., a flash card slot), an AC power port, and one or more network interfaces, such as a universal serial bus interface or an Ethernet port. For example, the monitor **209***b* can include a display **210***b* that can indicate a measurement for glucose, for example, in mg/dL. Other analytes and forms of display can also appear on the monitor **209***b*.

In addition, although a single sensor **201***b* with a single monitor **209***b* is shown, different combinations of sensors and device pairings can be implemented. For example, multiple sensors can be provided for a plurality of differing patient types or measurement sites or even patient fingers.

FIG. **2**C illustrates yet another example of monitoring device **200**C that can house the data collection system **100**. Like the monitoring device **200**B, the monitoring device **200**C includes a finger clip sensor **201***c* connected to a monitor **209***c* via a cable **212**. The cable **212** can have all of the features described above with respect to FIG. **2**B. The monitor **209***c* can include all of the features of the monitor **200**B described above. For example, the monitor **209***c* includes buttons **208***c* and a display **210***c*. The monitor **209***c* shown also includes straps **214***c* that allow the monitor **209***c* to be attached to a patient's limb or the like.

FIG. **2**D illustrates yet another example of monitoring device **200**D that can house the data collection system **100**. Like the monitoring devices **200**B and **200**C, the monitoring device **200**D includes a finger clip sensor **201***d* connected to a monitor **209***d* via a cable **212**. The cable **212** can have all of the features described above with respect to FIG. **2**B. In addition to having some or all of the features described above with respect to FIGS. **2**B and **2**C, the monitoring device **200**D includes an optional universal serial bus (USB) port **216** and an Ethernet port **218**. The USB port **216** and the Ethernet port **218** can be used, for example, to transfer information between the monitor **209***d* and a computer (not shown) via a cable. Software stored on the computer can provide functionality for a user to, for example, view physiological data and trends, adjust settings and download firmware updates to the monitor **209***b*, and perform a variety of other functions. The USB port **216** and the Ethernet port **218** can be included with the other monitoring devices **200**A, **200**B, and **200**C described above.

FIGS. **3**A through **3**C illustrate more detailed examples of embodiments of a sensor **301***a*. The sensor **301***a* shown can include all of the features of the sensors **100** and **200** described above.

Referring to FIG. **3**A, the sensor **301***a* in the depicted embodiment is a clothespin-shaped clip sensor that includes an enclosure **302***a* for receiving a patient's finger. The enclosure **302***a* is formed by an upper section or emitter shell **304***a*, which is pivotably connected with a lower section or detector shell **306***a*. The emitter shell **304***a* can be biased with the detector shell **306***a* to close together around a pivot point **303***a* and thereby sandwich finger tissue between the emitter and detector shells **304***a*, **306***a*.

In an embodiment, the pivot point **303***a* advantageously includes a pivot capable of adjusting the relationship between the emitter and detector shells **304***a*, **306***a* to effectively level the sections when applied to a tissue site. In another embodiment, the sensor **301***a* includes some or all features of the finger clip described in U.S. Publication No. 2006/0211924, incorporated above, such as a spring that causes finger clip forces to be distributed along the finger. Paragraphs [0096] through [0105], which describe this feature, are hereby specifically incorporated by reference.

The emitter shell **304***a* can position and house various emitter components of the sensor **301***a*. It can be constructed of reflective material (e.g., white silicone or plastic) and/or can be metallic or include metalicized plastic (e.g., including carbon and aluminum) to possibly serve as a heat sink. The emitter shell **304***a* can also include absorbing opaque mate-

rial, such as, for example, black or grey colored material, at various areas, such as on one or more flaps **307***a*, to reduce ambient light entering the sensor **301***a*.

The detector shell **306***a* can position and house one or more detector portions of the sensor **301***a*. The detector shell **306***a* can be constructed of reflective material, such as white silicone or plastic. As noted, such materials can increase the usable signal at a detector by forcing light back into the tissue and measurement site (see FIG. **1**). The detector shell **306***a* can also include absorbing opaque material at various areas, such as lower area **308***a*, to reduce ambient light entering the sensor **301***a*.

Referring to FIGS. **3**B and **3**C, an example of finger bed **310** is shown in the sensor **301***b*. The finger bed **310** includes a generally curved surface shaped generally to receive tissue, such as a human digit. The finger bed **310** includes one or more ridges or channels **314**. Each of the ridges **314** has a generally convex shape that can facilitate increasing traction or gripping of the patient's finger to the finger bed. Advantageously, the ridges **314** can improve the accuracy of spectroscopic analysis in certain embodiments by reducing noise that can result from a measurement site moving or shaking loose inside of the sensor **301***a*. The ridges **314** can be made from reflective or opaque materials in some embodiments to further increase SNR. In other implementations, other surface shapes can be used, such as, for example, generally flat, concave, or convex finger beds **310**.

Finger bed **310** can also include an embodiment of a tissue thickness adjuster or protrusion **305**. The protrusion **305** includes a measurement site contact area **370** (see FIG. **3**C) that can contact body tissue of a measurement site. The protrusion **305** can be removed from or integrated with the finger bed **310**. Interchangeable, different shaped protrusions **305** can also be provided, which can correspond to different finger shapes, characteristics, opacity, sizes, or the like.

Referring specifically to FIG. **3**C, the contact area **370** of the protrusion **305** can include openings or windows **320**, **321**, **322**, and **323**. When light from a measurement site passes through the windows **320**, **321**, **322**, and **323**, the light can reach one or more photodetectors (see FIG. **3**E). In an embodiment, the windows **320**, **321**, **322**, and **323** mirror specific detector placements layouts such that light can impinge through the protrusion **305** onto the photodetectors. Any number of windows **320**, **321**, **322**, and **323** can be employed in the protrusion **305** to allow light to pass from the measurement site to the photodetectors.

The windows **320**, **321**, **322**, and **323** can also include shielding, such as an embedded grid of wiring or a conductive glass coating, to reduce noise from ambient light or other electromagnetic noise. The windows **320**, **321**, **322**, and **323** can be made from materials, such as plastic or glass. In some embodiments, the windows **320**, **321**, **322**, and **323** can be constructed from conductive glass, such as indium tin oxide (ITO) coated glass. Conductive glass can be useful because its shielding is transparent, and thus allows for a larger aperture versus a window with an embedded grid of wiring. In addition, in certain embodiments, the conductive glass does not need openings in its shielding (since it is transparent), which enhances its shielding performance. For example, some embodiments that employ the conductive glass can attain up to an about 40% to about 50% greater signal than non-conductive glass with a shielding grid. In addition, in some embodiments, conductive glass can be useful for shielding noise from a greater variety of directions than non-conductive glass with a shielding grid.

Turning to FIG. **3**B, the sensor **301***a* can also include a shielding **315***a*, such as a metal cage, box, metal sheet, perforated metal sheet, a metal layer on a non-metal material, or the like. The shielding **315***a* is provided in the depicted embodiment below or embedded within the protrusion **305** to reduce noise. The shielding **315***a* can be constructed from a conductive material, such as copper. The shielding **315***a* can include one or more openings or windows (not shown). The windows can be made from glass or plastic to thereby allow light that has passed through the windows **320**, **321**, **322**, and **323** on an external surface of the protrusion **305** (see FIG. **3**C) to pass through to one or more photodetectors that can be enclosed or provided below (see FIG. **3**E).

In some embodiments, the shielding cage for shielding **315***a* can be constructed in a single manufactured component with or without the use of conductive glass. This form of construction may be useful in order to reduce costs of manufacture as well as assist in quality control of the components. Furthermore, the shielding cage can also be used to house various other components, such as sigma delta components for various embodiments of front end interfaces **108**.

In an embodiment, the photodetectors can be positioned within or directly beneath the protrusion **305** (see FIG. **3**E). In such cases, the mean optical path length from the emitters to the detectors can be reduced and the accuracy of blood analyte measurement can increase. For example, in one embodiment, a convex bump of about 1 mm to about 3 mm in height and about 10 mm^2 to about 60 mm^2 was found to help signal strength by about an order of magnitude versus other shapes. Of course other dimensions and sizes can be employed in other embodiments. Depending on the properties desired, the length, width, and height of the protrusion **305** can be selected. In making such determinations, consideration can be made of protrusion's **305** effect on blood flow at the measurement site and mean path length for optical radiation passing through openings **320**, **321**, **322**, and **323**. Patient comfort can also be considered in determining the size and shape of the protrusion.

In an embodiment, the protrusion **305** can include a pliant material, including soft plastic or rubber, which can somewhat conform to the shape of a measurement site. Pliant materials can improve patient comfort and tactility by conforming the measurement site contact area **370** to the measurement site. Additionally, pliant materials can minimize or reduce noise, such as ambient light. Alternatively, the protrusion **305** can be made from a rigid material, such as hard plastic or metal.

Rigid materials can improve measurement accuracy of a blood analyte by conforming the measurement site to the contact area **370**. The contact area **370** can be an ideal shape for improving accuracy or reducing noise. Selecting a material for the protrusion **305** can include consideration of materials that do not significantly alter blood flow at the measurement site. The protrusion **305** and the contact area **370** can include a combination of materials with various characteristics.

The contact area **370** serves as a contact surface for the measurement site. For example, in some embodiments, the contact area **370** can be shaped for contact with a patient's finger. Accordingly, the contact area **370** can be sized and shaped for different sizes of fingers. The contact area **370** can be constructed of different materials for reflective purposes as well as for the comfort of the patient. For example,

the contact area **370** can be constructed from materials having various hardness and textures, such as plastic, gel, foam, and the like.

The formulas and analysis that follow with respect to FIG. **5** provide insight into how selecting these variables can alter transmittance and intensity gain of optical radiation that has been applied to the measurement site. These examples do not limit the scope of this disclosure.

Referring to FIG. **5**, a plot **500** is shown that illustrates examples of effects of embodiments of the protrusion **305** on the SNR at various wavelengths of light. As described above, the protrusion **305** can assist in conforming the tissue and effectively reduce its mean path length. In some instances, this effect by the protrusion **305** can have significant impact on increasing the SNR.

According to the Beer Lambert law, a transmittance of light (I) can be expressed as follows: $I=I_o*e^{-m*b*c}$, where I_o is the initial power of light being transmitted, m is the path length traveled by the light, and the component "b*c" corresponds to the bulk absorption of the light at a specific wavelength of light. For light at about 1600 nm to about 1700 nm, for example, the bulk absorption component is generally around 0.7 mm^{-1}. Assuming a typical finger thickness of about 12 mm and a mean path length of 20 mm due to tissue scattering, then $I=I_o*e^{(-20*0.7)}$.

In an embodiment where the protrusion **305** is a convex bump, the thickness of the finger can be reduced to 10 mm (from 12 mm) for some fingers and the effective light mean path is reduced to about 16.6 mm from 20 mm (see box **510**). This results in a new transmittance, $I_1=I_o*e^{(-16.6*0.7)}$. A curve for a typical finger (having a mean path length of 20 mm) across various wavelengths is shown in the plot **500** of FIG. **5**. The plot **500** illustrates potential effects of the protrusion **305** on the transmittance. As illustrated, comparing I and I_1 results in an intensity gain of $e^{(-16.6*0.7)}/e^{(-20*0.7)}$, which is about a 10 times increase for light in the about 1600 nm to about 1700 nm range. Such an increase can affect the SNR at which the sensor can operate. The foregoing gains can be due at least in part to the about 1600 nm to about 1700 nm range having high values in bulk absorptions (water, protein, and the like), e.g., about 0.7 mm^{-1}. The plot **500** also shows improvements in the visible/near-infrared range (about 600 nm to about 1300 nm).

Turning again to FIGS. **3**A through **3**C, an example heat sink **350***a* is also shown. The heat sink **350***a* can be attached to, or protrude from an outer surface of, the sensor **301***a*, thereby providing increased ability for various sensor components to dissipate excess heat. By being on the outer surface of the sensor **301***a* in certain embodiments, the heat sink **350***a* can be exposed to the air and thereby facilitate more efficient cooling. In an embodiment, one or more of the emitters (see FIG. **1**) generate sufficient heat that inclusion of the heat sink **350***a* can advantageously allows the sensor **301***a* to remain safely cooled. The heat sink **350***a* can include one or more materials that help dissipate heat, such as, for example, aluminum, steel, copper, carbon, combinations of the same, or the like. For example, in some embodiments, the emitter shell **304***a* can include a heat conducting material that is also readily and relatively inexpensively moldable into desired shapes and forms.

In some embodiments, the heat sink **350***a* includes metalicized plastic. The metalicized plastic can include aluminum and carbon, for example. The material can allow for improved thermal conductivity and diffusivity, which can increase commercial viability of the heat sink. In some embodiments, the material selected to construct the heat sink **350***a* can include a thermally conductive liquid crystalline polymer, such as CoolPoly® D5506, commercially available from Cool Polymers®, Inc. of Warwick, R.I. Such a material can be selected for its electrically non-conductive and dielectric properties so as, for example, to aid in electrical shielding. In an embodiment, the heat sink **350***a* provides improved heat transfer properties when the sensor **301***a* is active for short intervals of less than a full day's use. In an embodiment, the heat sink **350***a* can advantageously provide improved heat transfers in about three (3) to about four (4) minute intervals, for example, although a heat sink **350***a* can be selected that performs effectively in shorter or longer intervals.

Moreover, the heat sink **350***a* can have different shapes and configurations for aesthetic as well as for functional purposes. In an embodiment, the heat sink is configured to maximize heat dissipation, for example, by maximizing surface area. In an embodiment, the heat sink **350***a* is molded into a generally curved surface and includes one or more fins, undulations, grooves, or channels. The example heat sink **350***a* shown includes fins **351***a* (see FIG. **3**A).

An alternative shape of a sensor **301***b* and heat sink **350***b* is shown in FIG. **3**D. The sensor **301***b* can include some or all of the features of the sensor **301***a*. For example, the sensor **301***b* includes an enclosure **302***b* formed by an emitter shell **304***b* and a detector shell **306***b*, pivotably connected about a pivot **303***a*. The emitter shell **304***b* can also include absorbing opaque material on one or more flaps **307***b*, and the detector shell **306***a* can also include absorbing opaque material at various areas, such as lower area **308***b*.

However, the shape of the sensor **301***b* is different in this embodiment. In particular, the heat sink **350***b* includes comb protrusions **351***b*. The comb protrusions **351***b* are exposed to the air in a similar manner to the fins **351***a* of the heat sink **350***a*, thereby facilitating efficient cooling of the sensor **301***b*.

FIG. **3**E illustrates a more detailed example of a detector shell **306***b* of the sensor **301***b*. The features described with respect to the detector shell **306***b* can also be used with the detector shell **306***a* of the sensor **301***a*.

As shown, the detector shell **306***b* includes detectors **316**. The detectors **316** can have a predetermined spacing **340** from each other, or a spatial relationship among one another that results in a spatial configuration. This spatial configuration can purposefully create a variation of path lengths among detectors **316** and the emitter discussed above.

In the depicted embodiment, the detector shell **316** can hold multiple (e.g., two, three, four, etc.) photodiode arrays that are arranged in a two-dimensional grid pattern. Multiple photodiode arrays can also be useful to detect light piping (e.g., light that bypasses measurement site **102**). In the detector shell **316**, walls can be provided to separate the individual photodiode arrays to prevent or reduce mixing of light signals from distinct quadrants. In addition, the detector shell **316** can be covered by windows of transparent material, such as glass, plastic, or the like, to allow maximum or increased transmission of power light captured. In various embodiments, the transparent materials used can also be partially transparent or translucent or can otherwise pass some or all of the optical radiation passing through them. As noted, this window can include some shielding in the form of an embedded grid of wiring, or a conductive layer or coating.

As further illustrated by FIG. **3**E, the detectors **316** can have a spatial configuration of a grid. However, the detectors **316** can be arranged in other configurations that vary the path length. For example, the detectors **316** can be arranged in a linear array, a logarithmic array, a two-dimensional

array, a zig-zag pattern, or the like. Furthermore, any number of the detectors **316** can be employed in certain embodiments.

FIG. **3**F illustrates another embodiment of a sensor **301***f*. The sensor **301***f* can include some or all of the features of the sensor **301***a* of FIG. **3**A described above. For example, the sensor **301***f* includes an enclosure **302***f* formed by an upper section or emitter shell **304***f*, which is pivotably connected with a lower section or detector shell **306***f* around a pivot point **303***f*. The emitter shell **304***f* can also include absorbing opaque material on various areas, such as on one or more flaps **307***f*, to reduce ambient light entering the sensor **301***f*. The detector shell **306***f* can also include absorbing opaque material at various areas, such as a lower area **308***f*. The sensor **301***f* also includes a heat sink **350***f*, which includes fins **351***f*.

In addition to these features, the sensor **301***f* includes a flex circuit cover **360**, which can be made of plastic or another suitable material. The flex circuit cover **360** can cover and thereby protect a flex circuit (not shown) that extends from the emitter shell **304***f* to the detector shell **306***f*. An example of such a flex circuit is illustrated in U.S. Publication No. 2006/0211924, incorporated above (see FIG. **46** and associated description, which is hereby specifically incorporated by reference). The flex circuit cover **360** is shown in more detail below in FIG. **17**.

In addition, sensors **301***a-f* has extra length—extends to second joint on finger—Easier to place, harder to move due to cable, better for light piping

FIGS. **4**A through **4**C illustrate example arrangements of a protrusion **405**, which is an embodiment of the protrusion **305** described above. In an embodiment, the protrusion **405** can include a measurement site contact area **470**. The measurement site contact area **470** can include a surface that molds body tissue of a measurement site, such as a finger, into a flat or relatively flat surface.

The protrusion **405** can have dimensions that are suitable for a measurement site such as a patient's finger. As shown, the protrusion **405** can have a length **400**, a width **410**, and a height **430**. The length **400** can be from about 9 to about 11 millimeters, e.g., about 10 millimeters. The width **410** can be from about 7 to about 9 millimeters, e.g., about 8 millimeters. The height **430** can be from about 0.5 millimeters to about 3 millimeters, e.g., about 2 millimeters. In an embodiment, the dimensions **400**, **410**, and **430** can be selected such that the measurement site contact area **470** includes an area of about 80 square millimeters, although larger and smaller areas can be used for different sized tissue for an adult, an adolescent, or infant, or for other considerations.

The measurement site contact area **470** can also include differently shaped surfaces that conform the measurement site into different shapes. For example, the measurement site contact area **470** can be generally curved and/or convex with respect to the measurement site. The measurement site contact area **470** can be other shapes that reduce or even minimize air between the protrusion **405** and or the measurement site. Additionally, the surface pattern of the measurement site contact area **470** can vary from smooth to bumpy, e.g., to provide varying levels of grip.

In FIGS. **4**A and **4**C, openings or windows **420**, **421**, **422**, and **423** can include a wide variety of shapes and sizes, including for example, generally square, circular, triangular, or combinations thereof. The windows **420**, **421**, **422**, and **423** can be of non-uniform shapes and sizes. As shown, the windows **420**, **421**, **422**, and **423** can be evenly spaced out in a grid like arrangement. Other arrangements or patterns of arranging the windows **420**, **421**, **422**, and **423** are possible. For example, the windows **420**, **421**, **422**, and **423** can be placed in a triangular, circular, or linear arrangement. In some embodiments, the windows **420**, **421**, **422**, and **423** can be placed at different heights with respect to the finger bed **310** of FIG. **3**. The windows **420**, **421**, **422**, and **423** can also mimic or approximately mimic a configuration of, or even house, a plurality of detectors.

FIGS. **6**A through **6**D illustrate another embodiment of a protrusion **605** that can be used as the tissue shaper **105** described above or in place of the protrusions **305**, **405** described above. The depicted protrusion **605** is a partially cylindrical lens having a partial cylinder **608** and an extension **610**. The partial cylinder **608** can be a half cylinder in some embodiments; however, a smaller or greater portion than half of a cylinder can be used. Advantageously, in certain embodiments, the partially cylindrical protrusion **605** focuses light onto a smaller area, such that fewer detectors can be used to detect the light attenuated by a measurement site.

FIG. **6**A illustrates a perspective view of the partially cylindrical protrusion **605**. FIG. **6**B illustrates a front elevation view of the partially cylindrical protrusion **605**. FIG. **6**C illustrates a side view of the partially cylindrical protrusion **605**. FIG. **6**D illustrates a top view of the partially cylindrical protrusion **605**.

Advantageously, in certain embodiments, placing the partially cylindrical protrusion **605** over the photodiodes in any of the sensors described above adds multiple benefits to any of the sensors described above. In one embodiment, the partially cylindrical protrusion **605** penetrates into the tissue and reduces the path length of the light traveling in the tissue, similar to the protrusions described above.

The partially cylindrical protrusion **605** can also collect light from a large surface and focus down the light to a smaller area. As a result, in certain embodiments, signal strength per area of the photodiode can be increased. The partially cylindrical protrusion **605** can therefore facilitate a lower cost sensor because, in certain embodiments, less photodiode area can be used to obtain the same signal strength. Less photodiode area can be realized by using smaller photodiodes or fewer photodiodes (see, e.g., FIG. **14**). If fewer or smaller photodiodes are used, the partially cylindrical protrusion **605** can also facilitate an improved SNR of the sensor because fewer or smaller photodiodes can have less dark current.

The dimensions of the partially cylindrical protrusion **605** can vary based on, for instance, a number of photodiodes used with the sensor. Referring to FIG. **6**C, the overall height of the partially cylindrical protrusion **605** (measurement "a") in some implementations is about 1 to about 3 mm. A height in this range can allow the partially cylindrical protrusion **605** to penetrate into the pad of the finger or other tissue and reduce the distance that light travels through the tissue. Other heights, however, of the partially cylindrical protrusion **605** can also accomplish this objective. For example, the chosen height of the partially cylindrical protrusion **605** can be selected based on the size of the measurement site, whether the patient is an adult or child, and so on. In an embodiment, the height of the protrusion **605** is chosen to provide as much tissue thickness reduction as possible while reducing or preventing occlusion of blood vessels in the tissue.

Referring to FIG. **6**D, the width of the partially cylindrical protrusion **605** (measurement "b") can be about 3 to about 5 mm. In one embodiment, the width is about 4 mm. In one embodiment, a width in this range provides good penetration

of the partially cylindrical protrusion **605** into the tissue to reduce the path length of the light. Other widths, however, of the partially cylindrical protrusion **605** can also accomplish this objective. For example, the width of the partially cylindrical protrusion **605** can vary based on the size of the measurement site, whether the patient is an adult or child, and so on. In addition, the length of the protrusion **605** could be about 10 mm, or about 8 mm to about 12 mm, or smaller than 8 mm or greater than 12 mm.

In certain embodiments, the focal length (f) for the partially cylindrical protrusion **605** can be expressed as:

$$f = \frac{R}{n-1},$$

where is the radius of curvature of the partial cylinder **608** and n is the index of refraction of the material used. In certain embodiments, the radius of curvature can be between about 1.5 mm and about 2 mm. In another embodiment, the partially cylindrical protrusion **605** can include a material, such as nBK7 glass, with an index of refraction of around 1.5 at 1300 nm, which can provide focal lengths of between about 3 mm and about 4 mm.

A partially cylindrical protrusion **605** having a material with a higher index of refraction such as nSF11 glass (e.g., n=1.75 at 1300 nm) can provide a shorter focal length and possibly a smaller photodiode chip, but can also cause higher reflections due to the index of refraction mismatch with air. Many types of glass or plastic can be used with index of refraction values ranging from, for example, about 1.4 to about 1.9. The index of refraction of the material of the protrusion **605** can be chosen to improve or optimize the light focusing properties of the protrusion **605**. A plastic partially cylindrical protrusion **605** could provide the cheapest option in high volumes but can also have some undesired light absorption peaks at wavelengths higher than 1500 nm. Other focal lengths and materials having different indices of refraction can be used for the partially cylindrical protrusion **605**.

Placing a photodiode at a given distance below the partially cylindrical protrusion **605** can facilitate capturing some or all of the light traveling perpendicular to the lens within the active area of the photodiode (see FIG. **14**). Different sizes of the partially cylindrical protrusion **605** can use different sizes of photodiodes. The extension **610** added onto the bottom of the partial cylinder **608** is used in certain embodiments to increase the height of the partially cylindrical protrusion **605**. In an embodiment, the added height is such that the photodiodes are at or are approximately at the focal length of the partially cylindrical protrusion **605**. In an embodiment, the added height provides for greater thinning of the measurement site. In an embodiment, the added height assists in deflecting light piped through the sensor. This is because light piped around the sensor passes through the side walls of the added height without being directed toward the detectors. The extension **610** can also further facilitate the protrusion **605** increasing or maximizing the amount of light that is provided to the detectors. In some embodiments, the extension **610** can be omitted.

FIG. **6**E illustrates another view of the sensor **301***f* of FIG. **3**F, which includes an embodiment of a partially cylindrical protrusion **605***b*. Like the sensor **301**A shown in FIGS. **3**B and **3**C, the sensor **301***f* includes a finger bed **310***f*. The finger bed **310***f* includes a generally curved surface shaped generally to receive tissue, such as a human digit. The finger bed **310***f* also includes the ridges or channels **314** described above with respect to FIGS. **3**B and **3**C.

The example of finger bed **310***f* shown also includes the protrusion **605***b*, which includes the features of the protrusion **605** described above. In addition, the protrusion **605***b* also includes chamfered edges **607** on each end to provide a more comfortable surface for a finger to slide across (see also FIG. **14**D). In another embodiment, the protrusion **605***b* could instead include a single chamfered edge **607** proximal to the ridges **314**. In another embodiment, one or both of the chamfered edges **607** could be rounded.

The protrusion **605***b* also includes a measurement site contact area **670** that can contact body tissue of a measurement site. The protrusion **605***b* can be removed from or integrated with the finger bed **310***f*. Interchangeable, differently shaped protrusions **605***b* can also be provided, which can correspond to different finger shapes, characteristics, opacity, sizes, or the like.

FIGS. **7**A and **7**B illustrate block diagrams of sensors **701** that include example arrangements of conductive glass or conductive coated glass for shielding. Advantageously, in certain embodiments, the shielding can provide increased SNR. The features of the sensors **701** can be implemented with any of the sensors **101**, **201**, **301** described above. Although not shown, the partially cylindrical protrusion **605** of FIG. **6** can also be used with the sensors **701** in certain embodiments.

For example, referring specifically to FIG. **7**A, the sensor **701***a* includes an emitter housing **704***a* and a detector housing **706**. The emitter housing **704***a* includes LEDs **104**. The detector housing **706***a* includes a tissue bed **710***a* with an opening or window **703***a*, the conductive glass **730***a*, and one or more photodiodes for detectors **106** provided on a submount **707***a*.

During operation, a finger **102** can be placed on the tissue bed **710***a* and optical radiation can be emitted from the LEDs **104**. Light can then be attenuated as it passes through or is reflected from the tissue of the finger **102**. The attenuated light can then pass through the opening **703***a* in the tissue bed **710***a*. Based on the received light, the detectors **106** can provide a detector signal **107**, for example, to the front end interface **108** (see FIG. **1**).

In the depicted embodiment, the conductive glass **730** is provided in the opening **703**. The conductive glass **730** can thus not only permit light from the finger to pass to the detectors **106**, but it can also supplement the shielding of the detectors **106** from noise. The conductive glass **730** can include a stack or set of layers. In FIG. **7**A, the conductive glass **730***a* is shown having a glass layer **731** proximate the finger **102** and a conductive layer **733** electrically coupled to the shielding **790***a*.

In an embodiment, the conductive glass **730***a* can be coated with a conductive, transparent or partially transparent material, such as a thin film of indium tin oxide (ITO). To supplement electrical shielding effects of a shielding enclosure **790***a*, the conductive glass **730***a* can be electrically coupled to the shielding enclosure **790***a*. The conductive glass **730***a* can be electrically coupled to the shielding **704***a* based on direct contact or via other connection devices, such as a wire or another component.

The shielding enclosure **790***a* can be provided to encompass the detectors **106** to reduce or prevent noise. For example, the shielding enclosure **790***a* can be constructed from a conductive material, such as copper, in the form of a metal cage. The shielding or enclosure a can include an opaque material to not only reduce electrical noise, but also ambient optical noise.

In some embodiments, the shielding enclosure **790***a* can be constructed in a single manufactured component with or without the use of conductive glass. This form of construction may be useful in order to reduce costs of manufacture as well as assist in quality control of the components. Furthermore, the shielding enclosure **790***a* can also be used to house various other components, such as sigma delta components for various embodiments of front end interfaces **108**.

Referring to FIG. **7**B, another block diagram of an example sensor **701***b* is shown. A tissue bed **710***b* of the sensor **701***b* includes a protrusion **705***b*, which is in the form of a convex bump. The protrusion **705***b* can include all of the features of the protrusions or tissue shaping materials described above. For example, the protrusion **705***b* includes a contact area **370** that comes in contact with the finger **102** and which can include one or more openings **703***b*. One or more components of conductive glass **730***b* can be provided in the openings **703**. For example, in an embodiment, each of the openings **703** can include a separate window of the conductive glass **730***b*. In an embodiment, a single piece of the conductive glass **730***b* can used for some or all of the openings **703***b*. The conductive glass **730***b* is smaller than the conductive glass **730***a* in this particular embodiment.

A shielding enclosure **790***b* is also provided, which can have all the features of the shielding enclosure **790***a*. The shielding enclosure **790***b* is smaller than the shielding enclosure **790***a*; however, a variety of sizes can be selected for the shielding enclosures **790**.

In some embodiments, the shielding enclosure **790***b* can be constructed in a single manufactured component with or without the use of conductive glass. This form of construction may be useful in order to reduce costs of manufacture as well as assist in quality control of the components. Furthermore, the shielding enclosure **790***b* can also be used to house various other components, such as sigma delta components for various embodiments of front end interfaces **108**.

FIGS. **8**A through **8**D illustrate a perspective view, side views, and a bottom elevation view of the conductive glass described above with respect to the sensors **701***a*, **701***b*. As shown in the perspective view of FIG. **8**A and side view of FIG. **8**B, the conductive glass **730** includes the electrically conductive material **733** described above as a coating on the glass layer **731** described above to form a stack. In an embodiment where the electrically conductive material **733** includes indium tin oxide, surface resistivity of the electrically conductive material **733** can range approximately from 30 ohms per square inch to 500 ohms per square inch, or approximately 30, 200, or 500 ohms per square inch. As would be understood by a person of skill in the art from the present disclosure, other resistivities can also be used which are less than 30 ohms or more than 500 ohms. Other transparent, electrically conductive materials can be used as the material **733**.

Although the conductive material **733** is shown spread over the surface of the glass layer **731**, the conductive material **733** can be patterned or provided on selected portions of the glass layer **731**. Furthermore, the conductive material **733** can have uniform or varying thickness depending on a desired transmission of light, a desired shielding effect, and other considerations.

In FIG. **8**C, a side view of a conductive glass **830***a* is shown to illustrate an embodiment where the electrically conductive material **733** is provided as an internal layer between two glass layers **731**, **835**. Various combinations of integrating electrically conductive material **733** with glass are possible. For example, the electrically conductive material **733** can be a layer within a stack of layers. This stack of layers can include one or more layers of glass **731**, **835**, as well as one or more layers of conductive material **733**. The stack can include other layers of materials to achieve desired characteristics.

In FIG. **8**D, a bottom perspective view is shown to illustrate an embodiment where a conductive glass **830***b* can include conductive material **837** that occupies or covers a portion of a glass layer **839**. This embodiment can be useful, for example, to create individual, shielded windows for detectors **106**, such as those shown in FIG. **3**C. The conductive material **837** can be patterned to include an area **838** to allow light to pass to detectors **106** and one or more strips **841** to couple to the shielding **704** of FIG. **7**.

Other configurations and patterns for the conductive material can be used in certain embodiments, such as, for example, a conductive coating lining periphery edges, a conductive coating outlaid in a pattern including a grid or other pattern, a speckled conductive coating, coating outlaid in lines in either direction or diagonally, varied thicknesses from the center out or from the periphery in, or other suitable patterns or coatings that balance the shielding properties with transparency considerations.

FIG. **9** depicts an example graph **900** that illustrates comparative results obtained by an example sensor having components similar to those disclosed above with respect to FIGS. **7** and **8**. The graph **900** depicts the results of the percentage of transmission of varying wavelengths of light for different types of windows used in the sensors described above.

A line **915** on the graph **900** illustrates example light transmission of a window made from plain glass. As shown, the light transmission percentage of varying wavelengths of light is approximately 90% for a window made from plain glass. A line **920** on the graph **900** demonstrates an example light transmission percentage for an embodiment in which a window is made from glass having an ITO coating with a surface resistivity of 500 ohms per square inch. A line **925** on the graph **900** shows an example light transmission for an embodiment in which a window is made from glass that includes a coating of ITO oxide with a surface resistivity of 200 ohms per square inch. A line **930** on the graph **900** shows an example light transmission for an embodiment in which a window is made from glass that includes a coating of ITO oxide with a surface resistivity of 30 ohms per square inch.

The light transmission percentage for a window with currently available embedded wiring can have a light transmission percentage of approximately 70%. This lower percentage of light transmission can be due to the opacity of the wiring employed in a currently available window with wiring. Accordingly, certain embodiments of glass coatings described herein can employ, for example, ITO coatings with different surface resistivity depending on the desired light transmission, wavelengths of light used for measurement, desired shielding effect, and other criteria.

FIGS. **10**A through **10**B illustrate comparative noise floors of example implementations of the sensors described above. Noise can include optical noise from ambient light and electro-magnetic noise, for example, from surrounding electrical equipment. In FIG. **10**A, a graph **1000** depicts possible noise floors for different frequencies of noise for an embodiment in which one of the sensors described above included separate windows for four (4) detectors **106**. One or more of the windows included an embedded grid of wiring as a noise shield. Symbols **1030**-**1033** illustrate the

noise floor performance for this embodiment. As can be seen, the noise floor performance can vary for each of the openings and based on the frequency of the noise.

In FIG. **10**B, a graph **1050** depicts a noise floor for frequencies of noise **1070** for an embodiment in which the sensor included separate openings for four (4) detectors **106** and one or more windows that include an ITO coating. In this embodiment, a surface resistivity of the ITO used was about 500 ohms per square inch. Symbols **1080**-**1083** illustrate the noise floor performance for this embodiment. As can be seen, the noise floor performance for this embodiment can vary less for each of the openings and provide lower noise floors in comparison to the embodiment of FIG. **10**A.

FIG. **11**A illustrates an example structure for configuring the set of optical sources of the emitters described above. As shown, an emitter **104** can include a driver **1105**, a thermistor **1120**, a set of top-emitting LEDs **1102** for emitting red and/or infrared light, a set of side-emitting LEDs **1104** for emitting near infrared light, and a submount **1106**.

The thermistor **1120** can be provided to compensate for temperature variations. For example, the thermistor **1120** can be provided to allow for wavelength centroid and power drift of LEDs **1102** and **1104** due to heating. In addition, other thermistors can be employed, for example, to measure a temperature of a measurement site. The temperature can be displayed on a display device and used by a caregiver. Such a temperature can also be helpful in correcting for wavelength drift due to changes in water absorption, which can be temperature dependent, thereby providing more accurate data useful in detecting blood analytes like glucose. In addition, using a thermistor or other type of temperature sensitive device may be useful for detecting extreme temperatures at the measurement site that are too hot or too cold. The presence of low perfusion may also be detected, for example, when the finger of a patient has become too cold. Moreover, shifts in temperature at the measurement site can alter the absorption spectrum of water and other tissue in the measurement cite. A thermistor's temperature reading can be used to adjust for the variations in absorption spectrum changes in the measurement site.

The driver **1105** can provide pulses of current to the emitter **1104**. In an embodiment, the driver **1105** drives the emitter **1104** in a progressive fashion, for example, in an alternating manner based on a control signal from, for example, a processor (e.g., the processor **110**). For example, the driver **1105** can drive the emitter **1104** with a series of pulses to about 1 milliwatt (mW) for visible light to light at about 1300 nm and from about 40 mW to about 100 mW for light at about 1600 nm to about 1700 nm. However, a wide number of driving powers and driving methodologies can be used. The driver **1105** can be synchronized with other parts of the sensor and can minimize or reduce any jitter in the timing of pulses of optical radiation emitted from the emitter **1104**. In some embodiments, the driver **1105** is capable of driving the emitter **1104** to emit an optical radiation in a pattern that varies by less than about 10 parts-per-million; however other amounts of variation can be used.

The submount **1106** provides a support structure in certain embodiments for aligning the top-emitting LEDs **1102** and the side-emitting LEDs **1104** so that their optical radiation is transmitted generally towards the measurement site. In some embodiments, the submount **1106** is also constructed of aluminum nitride (AlN) or beryllium oxide (BEO) for heat dissipation, although other materials or combinations of materials suitable for the submount **1106** can be used.

FIG. **11**B illustrates a configuration of emitting optical radiation into a measurement site for measuring a blood constituent or analyte like glucose. In some embodiments, emitter **104** may be driven in a progressive fashion to minimize noise and increase SNR of sensor **101**. For example, emitter **104** may be driven based on a progression of power/current delivered to LEDs **1102** and **1104**.

In some embodiments, emitter **104** may be configured to emit pulses centered about 905 nm, about 1050 nm, about 1200 nm, about 1300 nm, about 1330 nm, about 1610 nm, about 1640 nm, and about 1665 nm. In another embodiment, the emitter **104** may emit optical radiation ranging from about 860 nm to about 950 nm, about 950 nm to about 1100 nm, about 1100 nm to about 1270 nm, about 1250 nm to about 1350 nm, about 1300 nm to about 1360 nm, and about 1590 nm to about 1700 nm. Of course, emitter **104** may be configured to transmit any of a variety of wavelengths of visible, or near-infrared optical radiation.

For purposes of illustration, FIG. **11**B shows a sequence of pulses of light at wavelengths of around 905 nm, around 1200 nm, around 1300 nm, and around 1330 nm from top emitting LEDs **1102**. FIG. **11**B also shows that emitter **104** may then emit pulses centered at around 1630 nm, around 1660 nm, and around 1615 nm from side emitting LEDs **1104**. Emitter **104** may be progressively driven at higher power/current. This progression may allow driver circuit **105** to stabilize in its operations, and thus, provide a more stable current/power to LEDs **1102** and **1104**.

For example, as shown in FIG. **11**B, the sequence of optical radiation pulses are shown having a logarithmic-like progression in power/current. In some embodiments, the timing of these pulses is based on a cycle of about 400 slots running at 48 kHz (e.g. each time slot may be approximately 0.02 ms or 20 microseconds). An artisan will recognize that term "slots" includes its ordinary meaning, which includes a time period that may also be expressed in terms of a frequency. In the example shown, pulses from top emitting LEDs **1102** may have a pulse width of about 40 time slots (e.g., about 0.8 ms) and an off period of about 4 time slots in between. In addition, pulses from side emitting LEDs **1104** (e.g., or a laser diode) may have a pulse width of about 60 time slots (e.g., about 1.25 ms) and a similar off period of about 4 time slots. A pause of about 70 time slots (e.g. 1.5 ms) may also be provided in order to allow driver circuit **1105** to stabilize after operating at higher current/power.

As shown in FIG. **11**B, top emitting LEDs **1102** may be initially driven with a power to approximately 1 mW at a current of about 20-100 mA. Power in these LEDs may also be modulated by using a filter or covering of black dye to reduce power output of LEDs. In this example, top emitting LEDs **1102** may be driven at approximately 0.02 to 0.08 mW. The sequence of the wavelengths may be based on the current requirements of top emitting LEDs **502** for that particular wavelength. Of course, in other embodiments, different wavelengths and sequences of wavelengths may be output from emitter **104**.

Subsequently, side emitting LEDs **1104** may be driven at higher powers, such as about 40-100 mW and higher currents of about 600-800 mA. This higher power may be employed in order to compensate for the higher opacity of tissue and water in measurement site **102** to these wavelengths. For example, as shown, pulses at about 1630 nm, about 1660 nm, and about 1615 nm may be output with progressively higher power, such as at about 40 mW, about 50 mW, and about 60 mW, respectively. In this embodiment, the order of wavelengths may be based on the optical characteristics of that wavelength in tissue as well as the

current needed to drive side emitting LEDs **1104**. For example, in this embodiment, the optical pulse at about 1615 nm is driven at the highest power due to its sensitivity in detecting analytes like glucose and the ability of light at this wavelength to penetrate tissue. Of course, different wavelengths and sequences of wavelengths may be output from emitter **104**.

As noted, this progression may be useful in some embodiments because it allows the circuitry of driver circuit **1105** to stabilize its power delivery to LEDs **1102** and **1104**. Driver circuit **1105** may be allowed to stabilize based on the duty cycle of the pulses or, for example, by configuring a variable waiting period to allow for stabilization of driver circuit **1105**. Of course, other variations in power/current and wavelength may also be employed in the present disclosure.

Modulation in the duty cycle of the individual pulses may also be useful because duty cycle can affect the signal noise ratio of the system **100**. That is, as the duty cycle is increased so may the signal to noise ratio.

Furthermore, as noted above, driver circuit **1105** may monitor temperatures of the LEDs **1102** and **1104** using the thermistor **1120** and adjust the output of LEDs **1102** and **1104** accordingly. Such a temperature may be to help sensor **101** correct for wavelength drift due to changes in water absorption, which can be temperature dependent.

FIG. **11**C illustrates another exemplary emitter that may be employed in the sensor according to an embodiment of the disclosure. As shown, the emitter **104** can include components mounted on a substrate **1108** and on submount **1106**. In particular, top-emitting LEDs **1102** for emitting red and/or infrared light may be mounted on substrate **1108**. Side emitting LEDS **1104** may be mounted on submount **1106**. As noted, side-emitting LEDs **1104** may be included in emitter **104** for emitting near infrared light.

As also shown, the sensor of FIG. **11**C may include a thermistor **1120**. As noted, the thermistor **1120** can be provided to compensate for temperature variations. The thermistor **1120** can be provided to allow for wavelength centroid and power drift of LEDs **1102** and **1104** due to heating. In addition, other thermistors (not shown) can be employed, for example, to measure a temperature of a measurement site. Such a temperature can be helpful in correcting for wavelength drift due to changes in water absorption, which can be temperature dependent, thereby providing more accurate data useful in detecting blood analytes like glucose.

In some embodiments, the emitter **104** may be implemented without the use of side emitting LEDs. For example, certain blood constituents, such as total hemoglobin, can be measured by embodiments of the disclosure without the use of side emitting LEDs. FIG. **11**D illustrates another exemplary emitter that may be employed in the sensor according to an embodiment of the disclosure. In particular, an emitter **104** that is configured for a blood constituent, such as total hemoglobin, is shown. The emitter **104** can include components mounted on a substrate **1108**. In particular, top-emitting LEDs **1102** for emitting red and/or infrared light may be mounted on substrate **1108**.

As also shown, the emitter of FIG. **11**D may include a thermistor **1120**. The thermistor **1120** can be provided to compensate for temperature variations. The thermistor **1120** can be provided to allow for wavelength centroid and power drift of LEDs **1102** due to heating.

FIG. **12**A illustrates a detector submount **1200** having photodiode detectors that are arranged in a grid pattern on the detector submount **1200** to capture light at different quadrants from a measurement site. One detector submount **1200** can be placed under each window of the sensors described above, or multiple windows can be placed over a single detector submount **1200**. The detector submount **1200** can also be used with the partially cylindrical protrusion **605** described above with respect to FIG. **6**.

The detectors include photodiode detectors **1-4** that are arranged in a grid pattern on the submount **1200** to capture light at different quadrants from the measurement site. As noted, other patterns of photodiodes, such as a linear row, or logarithmic row, can also be employed in certain embodiments.

As shown, the detectors **1-4** may have a predetermined spacing from each other, or spatial relationship among one another that result in a spatial configuration. This spatial configuration can be configured to purposefully create a variation of path lengths among detectors **106** and the point light source discussed above.

Detectors may hold multiple (e.g., two, three, four, etc.) photodiode arrays that are arranged in a two-dimensional grid pattern. Multiple photodiode arrays may also be useful to detect light piping (i.e., light that bypasses measurement site **102**). As shown, walls may separate the individual photodiode arrays to prevent mixing of light signals from distinct quadrants. In addition, as noted, the detectors may be covered by windows of transparent material, such as glass, plastic, etc., to allow maximum transmission of power light captured. As noted, this window may comprise some shielding in the form of an embedded grid of wiring, or a conductive layer or coating.

FIGS. **12**B through **12**D illustrate a simplified view of exemplary arrangements and spatial configurations of photodiodes for detectors **106**. As shown, detectors **106** may comprise photodiode detectors **1-4** that are arranged in a grid pattern on detector submount **1200** to capture light at different quadrants from measurement site **102**.

As noted, other patterns of photodiodes may also be employed in embodiments of the present disclosure, including, for example, stacked or other configurations recognizable to an artisan from the disclosure herein. For example, detectors **106** may be arranged in a linear array, a logarithmic array, a two-dimensional array, and the like. Furthermore, an artisan will recognize from the disclosure herein that any number of detectors **106** may be employed by embodiments of the present disclosure.

For example, as shown in FIG. **12**B, detectors **106** may comprise photodiode detectors **1-4** that are arranged in a substantially linear configuration on submount **1200**. In this embodiment shown, photodiode detectors **1-4** are substantially equally spaced apart (e.g., where the distance D is substantially the same between detectors **1-4**).

In FIG. **12**C, photodiode detectors **1-4** may be arranged in a substantially linear configuration on submount **1200**, but may employ a substantially progressive, substantially logarithmic, or substantially semi-logarithmic spacing (e.g., where distances D**1**>D**2**>D**3**). This arrangement or pattern may be useful for use on a patient's finger and where the thickness of the finger gradually increases.

In FIG. **12**D, a different substantially grid pattern on submount **1200** of photodiode detectors **1-4** is shown. As noted, other patterns of detectors may also be employed in embodiments of the present invention.

FIGS. **12**E through **12**H illustrate several embodiments of photodiodes that may be used in detectors **106**. As shown in these figures, a photodiode **1202** of detector **106** may comprise a plurality of active areas **1204**, These active areas

204 may be coupled together via a common cathode 1206 or anode 1208 in order to provide a larger effective detection area.

In particular, as shown in FIG. 12E, photodiode 1202 may comprise two (2) active areas 1204*a* and 1204*b*. In FIG. 12F, photodiode 1202 may comprise four (4) active areas 1204*c-f*. In FIG. 12G, photodiode 1202 may comprise three (3) active areas 1204*g-i*. In FIG. 12H, photodiode 1202 may comprise nine (9) active areas 1204*j-r*. The use of smaller active areas may be useful because smaller active areas can be easier to fabricate and can be fabricated with higher purity. However, one skilled in the art will recognize that various sizes of active areas may be employed in the photodiode 1202.

FIG. 13 illustrates an example multi-stream process 1300. The multi-stream process 1300 can be implemented by the data collection system 100 and/or by any of the sensors described above. As shown, a control signal from a signal processor 1310 controls a driver 1305. In response, an emitter 1304 generates a pulse sequence 1303 from its emitter (e.g., its LEDs) into a measurement site or sites 1302. As described above, in some embodiments, the pulse sequence 1303 is controlled to have a variation of about 10 parts per million or less. Of course, depending on the analyte desired, the tolerated variation in the pulse sequence 1303 can be greater (or smaller).

In response to the pulse sequence 1300, detectors 1 to n (n being an integer) in a detector 1306 capture optical radiation from the measurement site 1302 and provide respective streams of output signals. Each signal from one of detectors 1-*n* can be considered a stream having respective time slots corresponding to the optical pulses from emitter sets 1-*n* in the emitter 1304. Although n emitters and n detectors are shown, the number of emitters and detectors need not be the same in certain implementations.

A front end interface 1308 can accept these multiple streams from detectors 1-*n* and deliver one or more signals or composite signal(s) back to the signal processor 1310. A stream from the detectors 1-*n* can thus include measured light intensities corresponding to the light pulses emitted from the emitter 1304.

The signal processor 1310 can then perform various calculations to measure the amount of glucose and other analytes based on these multiple streams of signals. In order to help explain how the signal processor 1310 can measure analytes like glucose, a primer on the spectroscopy employed in these embodiments will now be provided.

Spectroscopy is premised upon the Beer-Lambert law. According to this law, the properties of a material, e.g., glucose present in a measurement site, can be deterministically calculated from the absorption of light traveling through the material. Specifically, there is a logarithmic relation between the transmission of light through a material and the concentration of a substance and also between the transmission and the length of the path traveled by the light. As noted, this relation is known as the Beer-Lambert law.

The Beer-Lambert law is usually written as:

Absorbance A=m*b*c, where:

m is the wavelength-dependent molar absorptivity coefficient (usually expressed in units of M^{-1} cm^{-1});

b is the mean path length; and

c is the analyte concentration (e.g., the desired parameter).

In spectroscopy, instruments attempt to obtain the analyte concentration (c) by relating absorbance (A) to transmittance (T). Transmittance is a proportional value defined as:

$T=I/I_o$, where:

I is the light intensity measured by the instrument from the measurement site; and

I_o is the initial light intensity from the emitter.

Absorbance (A) can be equated to the transmittance (T) by the equation:

$$A=-\log T$$

Therefore, substituting equations from above:

$$A=-\log(I/I_o)$$

In view of this relationship, spectroscopy thus relies on a proportional-based calculation of $-\log(I/I_o)$ and solving for analyte concentration (c).

Typically, in order to simplify the calculations, spectroscopy will use detectors that are at the same location in order to keep the path length (b) a fixed, known constant. In addition, spectroscopy will employ various mechanisms to definitively know the transmission power (I_o), such as a photodiode located at the light source. This architecture can be viewed as a single channel or single stream sensor, because the detectors are at a single location.

However, this scheme can encounter several difficulties in measuring analytes, such as glucose. This can be due to the high overlap of absorption of light by water at the wavelengths relevant to glucose as well as other factors, such as high self-noise of the components.

Embodiments of the present disclosure can employ a different approach that in part allows for the measurement of analytes like glucose. Some embodiments can employ a bulk, non-pulsatile measurement in order to confirm or validate a pulsatile measurement. In addition, both the non-pulsatile and pulsatile measurements can employ, among other things, the multi-stream operation described above in order to attain sufficient SNR. In particular, a single light source having multiple emitters can be used to transmit light to multiple detectors having a spatial configuration.

A single light source having multiple emitters can allow for a range of wavelengths of light to be used. For example, visible, infrared, and near infrared wavelengths can be employed. Varying powers of light intensity for different wavelengths can also be employed.

Secondly, the use of multiple-detectors in a spatial configuration allow for a bulk measurement to confirm or validate that the sensor is positioned correctly. This is because the multiple locations of the spatial configuration can provide, for example, topology information that indicates where the sensor has been positioned. Currently available sensors do not provide such information. For example, if the bulk measurement is within a predetermined range of values, then this can indicate that the sensor is positioned correctly in order to perform pulsatile measurements for analytes like glucose. If the bulk measurement is outside of a certain range or is an unexpected value, then this can indicate that the sensor should be adjusted, or that the pulsatile measurements can be processed differently to compensate, such as using a different calibration curve or adjusting a calibration curve. This feature and others allow the embodiments to achieve noise cancellation and noise reduction, which can be several times greater in magnitude that what is achievable by currently available technology.

In order to help illustrate aspects of the multi-stream measurement approach, the following example derivation is provided. Transmittance (T) can be expressed as:

$$T=e^{-m*b*c}$$

In terms of light intensity, this equation can also be rewritten as:

$$I/I_o = e^{-m*b*c}$$

Or, at a detector, the measured light (I) can be expressed as:

$$I = I_o * e^{-m*b*c}$$

As noted, in the present disclosure, multiple detectors (**1** to n) can be employed, which results in $I_1 \ldots I_n$ streams of measurements. Assuming each of these detectors have their own path lengths, $b_1 \ldots b_n$, from the light source, the measured light intensities can be expressed as:

$$I_n = I_o * e^{-m*b_n*c}$$

The measured light intensities at any two different detectors can be referenced to each other. For example:

$$I_1/I_n = (I_o * e^{-mb_1c})/(I_o * e^{-mb_nc})$$

As can be seen, the terms, I_o, cancel out and, based on exponent algebra, the equation can be rewritten as:

$$I_1/I_n = e^{-m(b_1-b_n)c}$$

From this equation, the analyte concentration (c) can now be derived from bulk signals $I_1 \ldots I_n$ and knowing the respective mean path lengths b_1 and b_n. This scheme also allows for the cancelling out of I_o, and thus, noise generated by the emitter **1304** can be cancelled out or reduced. In addition, since the scheme employs a mean path length difference, any changes in mean path length and topological variations from patient to patient are easily accounted. Furthermore, this bulk-measurement scheme can be extended across multiple wavelengths. This flexibility and other features allow embodiments of the present disclosure to measure blood analytes like glucose.

For example, as noted, the non-pulsatile, bulk measurements can be combined with pulsatile measurements to more accurately measure analytes like glucose. In particular, the non-pulsatile, bulk measurement can be used to confirm or validate the amount of glucose, protein, etc. in the pulsatile measurements taken at the tissue at the measurement site(s) **1302**. The pulsatile measurements can be used to measure the amount of glucose, hemoglobin, or the like that is present in the blood. Accordingly, these different measurements can be combined to thus determine analytes like blood glucose.

FIG. **14**A illustrates an embodiment of a detector submount **1400***a* positioned beneath the partially cylindrical protrusion **605** of FIG. **6** (or alternatively, the protrusion **605***b*). The detector submount **1400***a* includes two rows **1408***a* of detectors **1410***a*. The partially cylindrical protrusion **605** can facilitate reducing the number and/or size of detectors used in a sensor because the protrusion **605** can act as a lens that focuses light onto a smaller area.

To illustrate, in some sensors that do not include the partially cylindrical protrusion **605**, sixteen detectors can be used, including four rows of four detectors each. Multiple rows of detectors can be used to measure certain analytes, such as glucose or total hemoglobin, among others. Multiple rows of detectors can also be used to detect light piping (e.g., light that bypasses the measurement site). However, using more detectors in a sensor can add cost, complexity, and noise to the sensor.

Applying the partially cylindrical protrusion **605** to such a sensor, however, could reduce the number of detectors or rows of detectors used while still receiving the substantially same amount of light, due to the focusing properties of the protrusion **605** (see FIG. **14**B). This is the example situation illustrated in FIG. **14**—two rows **1408***a* of detectors **1410***a* are used instead of four. Advantageously, in certain embodiments, the resulting sensor can be more cost effective, have less complexity, and have an improved SNR, due to fewer and/or smaller photodiodes.

In other embodiments, using the partially cylindrical protrusion **605** can allow the number of detector rows to be reduced to one or three rows of four detectors. The number of detectors in each row can also be reduced. Alternatively, the number of rows might not be reduced but the size of the detectors can be reduced. Many other configurations of detector rows and sizes can also be provided.

FIG. **14**B depicts a front elevation view of the partially cylindrical protrusion **605** (or alternatively, the protrusion **605***b*) that illustrates how light from emitters (not shown) can be focused by the protrusion **605** onto detectors. The protrusion **605** is placed above a detector submount **1400***b* having one or more detectors **1410***b* disposed thereon. The submount **1400***b* can include any number of rows of detectors **1410**, although one row is shown.

Light, represented by rays **1420**, is emitted from the emitters onto the protrusion **605**. These light rays **1420** can be attenuated by body tissue (not shown). When the light rays **1420** enter the protrusion **605**, the protrusion **605** acts as a lens to refract the rays into rays **1422**. This refraction is caused in certain embodiments by the partially cylindrical shape of the protrusion **605**. The refraction causes the rays **1422** to be focused or substantially focused on the one or more detectors **1410***b*. Since the light is focused on a smaller area, a sensor including the protrusion **605** can include fewer detectors to capture the same amount of light compared with other sensors.

FIG. **14**C illustrates another embodiment of a detector submount **1400***c*, which can be disposed under the protrusion **605***b* (or alternatively, the protrusion **605**). The detector submount **1400***c* includes a single row **1408***c* of detectors **1410***c*. The detectors are electrically connected to conductors **1412***c*, which can be gold, silver, copper, or any other suitable conductive material.

The detector submount **1400***c* is shown positioned under the protrusion **605***b* in a detector subassembly **1450** illustrated in FIG. **14**D. A top-down view of the detector subassembly **1450** is also shown in FIG. **14**E. In the detector subassembly **1450**, a cylindrical housing **1430** is disposed on the submount **1400***c*. The cylindrical housing **1430** includes a transparent cover **1432**, upon which the protrusion **605***b* is disposed. Thus, as shown in FIG. **14**D, a gap **1434** exists between the detectors **1410***c* and the protrusion **605***b*. The height of this gap **1434** can be chosen to increase or maximize the amount of light that impinges on the detectors **1410***c*.

The cylindrical housing **1430** can be made of metal, plastic, or another suitable material. The transparent cover **1432** can be fabricated from glass or plastic, among other materials. The cylindrical housing **1430** can be attached to the submount **1400***c* at the same time or substantially the same time as the detectors **1410***c* to reduce manufacturing costs. A shape other than a cylinder can be selected for the housing **1430** in various embodiments.

In certain embodiments, the cylindrical housing **1430** (and transparent cover **1432**) forms an airtight or substantially airtight or hermetic seal with the submount **1400***c*. As a result, the cylindrical housing **1430** can protect the detectors **1410***c* and conductors **1412***c* from fluids and vapors that can cause corrosion. Advantageously, in certain embodiments, the cylindrical housing **1430** can protect the detectors **1410***c* and conductors **1412***c* more effectively than currently-available resin epoxies, which are sometimes applied to solder joints between conductors and detectors.

In embodiments where the cylindrical housing **1430** is at least partially made of metal, the cylindrical housing **1430** can provide noise shielding for the detectors **1410***c*. For example, the cylindrical housing **1430** can be soldered to a ground connection or ground plane on the submount **1400***c*, which allows the cylindrical housing **1430** to reduce noise. In another embodiment, the transparent cover **1432** can include a conductive material or conductive layer, such as conductive glass or plastic. The transparent cover **1432** can include any of the features of the noise shields **790** described above.

The protrusion **605***b* includes the chamfered edges **607** described above with respect to FIG. **6**E. These chamfered edges **607** can allow a patient to more comfortably slide a finger over the protrusion **605***b* when inserting the finger into the sensor **301***f*.

FIG. **14**F illustrates a portion of the detector shell **306***f*, which includes the detectors **1410***c* on the substrate **1400***c*. The substrate **1400***c* is enclosed by a shielding enclosure **1490**, which can include the features of the shielding enclosures **790***a*, **790***b* described above (see also FIG. **17**). The shielding enclosure **1490** can be made of metal. The shielding enclosure **1490** includes a window **1492***a* above the detectors **1410***c*, which allows light to be transmitted onto the detectors **1410***c*.

A noise shield **1403** is disposed above the shielding enclosure **1490**. The noise shield **1403**, in the depicted embodiment, includes a window **1492***a* corresponding to the window **1492***a*. Each of the windows **1492***a*, **1492***b* can include glass, plastic, or can be an opening without glass or plastic. In some embodiments, the windows **1492***a*, **1492***b* may be selected to have different sizes or shapes from each other.

The noise shield **1403** can include any of the features of the conductive glass described above. In the depicted embodiment, the noise shield **1403** extends about three-quarters of the length of the detector shell **306***f*. In other embodiments, the noise shield **1403** could be smaller or larger. The noise shield **1403** could, for instance, merely cover the detectors **1410***c*, the submount **1400***c*, or a portion thereof. The noise shield **1403** also includes a stop **1413** for positioning a measurement site within the sensor **301***f*. Advantageously, in certain embodiments, the noise shield **1403** can reduce noise caused by light piping.

A thermistor **1470** is also shown. The thermistor **1470** is attached to the submount **1400***c* and protrudes above the noise shield **1403**. As described above, the thermistor **1470** can be employed to measure a temperature of a measurement site. Such a temperature can be helpful in correcting for wavelength drift due to changes in water absorption, which can be temperature dependent, thereby providing more accurate data useful in detecting blood analytes like glucose.

In the depicted embodiment, the detectors **1410***c* are not enclosed in the cylindrical housing **1430**. In an alternative embodiment, the cylindrical housing **1430** encloses the detectors **1410***c* and is disposed under the noise shield **1403**. In another embodiment, the cylindrical housing **1430** encloses the detectors **1410***c* and the noise shield **1403** is not used. If both the cylindrical housing **1403** and the noise shield **1403** are used, either or both can have noise shielding features.

FIG. **14**G illustrates the detector shell **306***f* of FIG. **14**F, with the finger bed **310***f* disposed thereon. FIG. **14**H illustrates the detector shell **306***f* of FIG. **14**G, with the protrusion **605***b* disposed in the finger bed **310***f*.

FIG. **14**I illustrates a cutaway view of the sensor **301***f*. Not all features of the sensor **301***f* are shown, such as the protrusion **605***b*. Features shown include the emitter and detector shells **304***f*, **306***f*, the flaps **307***f*, the heat sink **350***f* and fins **351***f*, the finger bed **310***f*, and the noise shield **1403**.

In addition to these features, emitters **1404** are depicted in the emitter shell **304***f*. The emitters **1404** are disposed on a submount **1401**, which is connected to a circuit board **1419**. The emitters **1404** are also enclosed within a cylindrical housing **1480**. The cylindrical housing **1480** can include all of the features of the cylindrical housing **1430** described above. For example, the cylindrical housing **1480** can be made of metal, can be connected to a ground plane of the submount **1401** to provide noise shielding, and can include a transparent cover **1482**.

The cylindrical housing **1480** can also protect the emitters **1404** from fluids and vapors that can cause corrosion. Moreover, the cylindrical housing **1480** can provide a gap between the emitters **1404** and the measurement site (not shown), which can allow light from the emitters **1404** to even out or average out before reaching the measurement site.

The heat sink **350***f*, in addition to including the fins **351***f*, includes a protuberance **352***f* that extends down from the fins **351***f* and contacts the submount **1401**. The protuberance **352***f* can be connected to the submount **1401**, for example, with thermal paste or the like. The protuberance **352***f* can sink heat from the emitters **1404** and dissipate the heat via the fins **351***f*.

FIGS. **15**A and **15**B illustrate embodiments of sensor portions **1500**A, **1500**B that include alternative heat sink features to those described above. These features can be incorporated into any of the sensors described above. For example, any of the sensors above can be modified to use the heat sink features described below instead of or in addition to the heat sink features of the sensors described above.

The sensor portions **1500**A, **1500**B shown include LED emitters **1504**; however, for ease of illustration, the detectors have been omitted. The sensor portions **1500**A, **1500**B shown can be included, for example, in any of the emitter shells described above.

The LEDs **1504** of the sensor portions **1500**A, **1500**B are connected to a substrate or submount **1502**. The submount **1502** can be used in place of any of the submounts described above. The submount **1502** can be a non-electrically conducting material made of any of a variety of materials, such as ceramic, glass, or the like. A cable **1512** is attached to the submount **1502** and includes electrical wiring **1514**, such as twisted wires and the like, for communicating with the LEDs **1504**. The cable **1512** can correspond to the cables **212** described above.

Although not shown, the cable **1512** can also include electrical connections to a detector. Only a portion of the cable **1512** is shown for clarity. The depicted embodiment of the cable **1512** includes an outer jacket **1510** and a conductive shield **1506** disposed within the outer jacket **1510**. The conductive shield **1506** can be a ground shield or the like that is made of a metal such as braided copper or aluminum. The conductive shield **1506** or a portion of the conductive shield **1506** can be electrically connected to the submount **1502** and can reduce noise in the signal generated by the sensor **1500**A, **1500**B by reducing RF coupling with the wires **1514**. In alternative embodiments, the cable **1512** does not have a conductive shield. For example, the cable **1512** could be a twisted pair cable or the like, with one wire of the twisted pair used as a heat sink.

Referring specifically to FIG. **15**A, in certain embodiments, the conductive shield **1506** can act as a heat sink for the LEDs **1504** by absorbing thermal energy from the LEDs **1504** and/or the submount **1502**. An optional heat insulator **1520** in communication with the submount **1502** can also assist with directing heat toward the conductive shield **1506**. The heat insulator **1520** can be made of plastic or another suitable material. Advantageously, using the conductive shield **1506** in the cable **1512** as a heat sink can, in certain embodiments, reduce cost for the sensor.

Referring to FIG. **15**B, the conductive shield **1506** can be attached to both the submount **1502** and to a heat sink layer **1530** sandwiched between the submount **1502** and the optional insulator **1520**. Together, the heat sink layer **1530** and the conductive shield **1506** in the cable **1512** can absorb at least part of the thermal energy from the LEDs and/or the submount **1502**.

FIGS. **15**C and **15**D illustrate implementations of a sensor portion **1500**C that includes the heat sink features of the sensor portion **1500**A described above with respect to FIG. **15**A. The sensor portion **1500**C includes the features of the sensor portion **1500**A, except that the optional insulator **1520** is not shown. FIG. **15**D is a side cutaway view of the sensor portion **1500**C that shows the emitters **1504**.

The cable **1512** includes the outer jacket **1510** and the conductive shield **1506**. The conductive shield **1506** is soldered to the submount **1502**, and the solder joint **1561** is shown. In some embodiments, a larger solder joint **1561** can assist with removing heat more rapidly from the emitters **1504**. Various connections **1563** between the submount **1502** and a circuit board **1519** are shown. In addition, a cylindrical housing **1580**, corresponding to the cylindrical housing **1480** of FIG. **14**I, is shown protruding through the circuit board **1519**. The emitters **1504** are enclosed in the cylindrical housing **1580**.

FIGS. **15**E and **15**F illustrate implementations of a sensor portion **1500**E that includes the heat sink features of the sensor portion **1500**B described above with respect to FIG. **15**B. The sensor portion **1500**E includes the heat sink layer **1530**. The heat sink layer **1530** can be a metal plate, such as a copper plate or the like. The optional insulator **1520** is not shown. FIG. **15**F is a side cutaway view of the sensor portion **1500**E that shows the emitters **1504**.

In the depicted embodiment, the conductive shield **1506** of the cable **1512** is soldered to the heat sink layer **1530** instead of the submount **1502**. The solder joint **1565** is shown. In some embodiments, a larger solder joint **1565** can assist with removing heat more rapidly from the emitters **1504**. Various connections **1563** between the submount **1502** and a circuit board **1519** are shown. In addition, the cylindrical housing **1580** is shown protruding through the circuit board **1519**. The emitters **1504** are enclosed in the cylindrical housing **1580**.

FIGS. **15**G and **15**H illustrate embodiments of connector features that can be used with any of the sensors described above with respect to FIGS. **1** through **15**F. Referring to FIG. **15**G, the circuit board **1519** includes a female connector **1575** that mates with a male connector **1577** connected to a daughter board **1587**. The daughter board **1587** includes connections to the electrical wiring **1514** of the cable **1512**. The connected boards **1519**, **1587** are shown in FIG. **15**H. Also shown is a hole **1573** that can receive the cylindrical housing **1580** described above.

Advantageously, in certain embodiments, using a daughter board **1587** to connect to the circuit board **1519** can enable connections to be made more easily to the circuit board **1519**. In addition, using separate boards can be easier to manufacture than a single circuit board **1519** with all connections soldered to the circuit board **1519**.

FIG. **15**I illustrates an exemplary architecture for front-end interface **108** as a transimpedance-based front-end. As noted, front-end interfaces **108** provide an interface that adapts the output of detectors **106** into a form that can be handled by signal processor **110**. As shown in this figure, sensor **101** and front-end interfaces **108** may be integrated together as a single component, such as an integrated circuit. Of course, one skilled in the art will recognize that sensor **101** and front end interfaces **108** may comprise multiple components or circuits that are coupled together.

Front-end interfaces **108** may be implemented using transimpedance amplifiers that are coupled to analog to digital converters in a sigma delta converter. In some embodiments, a programmable gain amplifier (PGA) can be used in combination with the transimpedance-based front-ends. For example, the output of a transimpedance-based front-end may be output to a sigma-delta ADC that comprises a PGA. A PGA may be useful in order to provide another level of amplification and control of the stream of signals from detectors **106**. The PGA may be an integrated circuit or built from a set of micro-relays. Alternatively, the PGA and ADC components in converter **900** may be integrated with the transimpedance-based front-end in sensor **101**.

Due to the low-noise requirements for measuring blood analytes like glucose and the challenge of using multiple photodiodes in detector **106**, the applicants developed a noise model to assist in configuring front-end **108**. Conventionally, those skilled in the art have focused on optimizing the impedance of the transimpedance amplifiers to minimize noise.

However, the following noise model was discovered by the applicants:

$$\text{Noise} = \sqrt{aR + bR^2},$$

aR is characteristic of the impedance of the transimpedance amplifier; and

bR^2 is characteristic of the impedance of the photodiodes in detector and the number of photodiodes in detector **106**.

The foregoing noise model was found to be helpful at least in part due to the high SNR required to measure analytes like glucose. However, the foregoing noise model was not previously recognized by artisans at least in part because, in conventional devices, the major contributor to noise was generally believed to originate from the emitter or the LEDs. Therefore, artisans have generally continued to focus on reducing noise at the emitter.

However, for analytes like glucose, the discovered noise model revealed that one of the major contributors to noise was generated by the photodiodes. In addition, the amount of noise varied based on the number of photodiodes coupled to a transimpedance amplifier. Accordingly, combinations of various photodiodes from different manufacturers, different impedance values with the transimpedance amplifiers, and different numbers of photodiodes were tested as possible embodiments.

In some embodiments, different combinations of transimpedance to photodiodes may be used. For example, detectors **1-4** (as shown, e.g., in FIG. **12**A) may each comprise four photodiodes. In some embodiments, each detector of four photodiodes may be coupled to one or more transimpedance

amplifiers. The configuration of these amplifiers may be set according to the model shown in FIG. **15**J.

Alternatively, each of the photodiodes may be coupled to its own respective transimpedance amplifier. For example, transimpedance amplifiers may be implemented as integrated circuits on the same circuit board as detectors **1**-**4**. In this embodiment, the transimpedance amplifiers may be grouped into an averaging (or summing) circuit, which are known to those skilled in the art, in order to provide an output stream from the detector. The use of a summing amplifier to combine outputs from several transimpedance amplifiers into a single, analog signal may be helpful in improving the SNR relative to what is obtainable from a single transimpedance amplifier. The configuration of the transimpedance amplifiers in this setting may also be set according to the model shown in FIG. **15**J.

As yet another alternative, as noted above with respect to FIGS. **12**E through **12**H, the photodiodes in detectors **106** may comprise multiple active areas that are grouped together. In some embodiments, each of these active areas may be provided its own respective transimpedance. This form of pairing may allow a transimpedance amplifier to be better matched to the characteristics of its corresponding photodiode or active area of a photodiode.

As noted, FIG. **15**J illustrates an exemplary noise model that may be useful in configuring transimpedance amplifiers. As shown, for a given number of photodiodes and a desired SNR, an optimal impedance value for a transimpedance amplifier could be determined.

For example, an exemplary "4 PD per stream" sensor **1502** is shown where detector **106** comprises four photodiodes **1502**. The photodiodes **1502** are coupled to a single transimpedance amplifier **1504** to produce an output stream **1506**. In this example, the transimpedance amplifier comprises 10 MO resistors **1508** and **1510**. Thus, output stream **1506** is produced from the four photodiodes (PD) **1502**. As shown in the graph of FIG. **15**J, the model indicates that resistance values of about 10 MO may provide an acceptable SNR for analytes like glucose.

However, as a comparison, an exemplary "1 PD per stream" sensor **1512** is also shown in FIG. **15**J. In particular, sensor **1512** may comprise a plurality of detectors **106** that each comprises a single photodiode **1514**. In addition, as shown for this example configuration, each of photodiodes **1514** may be coupled to respective transimpedance amplifiers **1516**, e.g., 1 PD per stream. Transimpedance amplifiers are shown having 40 MO resistors **1518**. As also shown in the graph of FIG. **15**J, the model illustrates that resistance values of 40 MO for resistors **1518** may serve as an alternative to the 4 photodiode per stream architecture of sensor **1502** described above and yet still provide an equivalent SNR.

Moreover, the discovered noise model also indicates that utilizing a 1 photodiode per stream architecture like that in sensor **1512** may provide enhanced performance because each of transimpedance amplifiers **1516** can be tuned or optimized to its respective photodiodes **1518**. In some embodiments, an averaging component **1520** may also be used to help cancel or reduce noise across photodiodes **1518**.

For purposes of illustration, FIG. **15**K shows different architectures (e.g., four PD per stream and one PD per stream) for various embodiments of a sensor and how components of the sensor may be laid out on a circuit board or substrate. For example, sensor **1522** may comprise a "4 PD per stream" architecture on a submount **700** in which each detector **106** comprises four (4) photodiodes **1524**. As shown for sensor **1522**, the output of each set of four photodiodes **1524** is then aggregated into a single transimpedance amplifier **1526** to produce a signal.

As another example, a sensor **1528** may comprise a "1 PD per stream" architecture on submount **700** in which each detector **106** comprises four (4) photodiodes **1530**. In sensor **1528**, each individual photodiode **1530** is coupled to a respective transimpedance amplifier **1532**. The output of the amplifiers **1532** may then be aggregated into averaging circuit **1520** to produce a signal.

As noted previously, one skilled in the art will recognize that the photodiodes and detectors may be arranged in different fashions to optimize the detected light. For example, sensor **1534** illustrates an exemplary "4 PD per stream" sensor in which the detectors **106** comprise photodiodes **1536** arranged in a linear fashion. Likewise, sensor **1538** illustrates an exemplary "1 PD per stream" sensor in which the detectors comprise photodiodes **1540** arranged in a linear fashion.

Alternatively, sensor **1542** illustrates an exemplary "4 PD per stream" sensor in which the detectors **106** comprise photodiodes **1544** arranged in a two-dimensional pattern, such as a zig-zag pattern. Sensor **1546** illustrates an exemplary "1 PD per stream" sensor in which the detectors comprise photodiodes **1548** also arranged in a zig-zag pattern.

FIG. **15**L illustrates an exemplary architecture for a switched-capacitor-based front-end. As shown, front-end interfaces **108** may be implemented using switched capacitor circuits and any number of front-end interfaces **108** may be implemented. The output of these switched capacitor circuits may then be provided to a digital interface **1000** and signal processor **110**. Switched capacitor circuits may be useful in system **100** for their resistor free design and analog averaging properties. In particular, the switched capacitor circuitry provides for analog averaging of the signal that allows for a lower smaller sampling rate (e.g., 2 KHz sampling for analog versus 48 KHz sampling for digital designs) than similar digital designs. In some embodiments, the switched capacitor architecture in front end interfaces **108** may provide a similar or equivalent SNR to other front end designs, such as a sigma delta architecture. In addition, a switched capacitor design in front end interfaces **108** may require less computational power by signal processor **110** to perform the same amount of decimation to obtain the same SNR.

FIGS. **16**A and **16**B illustrate embodiments of disposable optical sensors **1600**. In an embodiment, any of the features described above, such as protrusion, shielding, and/or heat sink features, can be incorporated into the disposable sensors **1600** shown. For instance, the sensors **1600** can be used as the sensors **101** in the system **100** described above with respect to FIG. **1**. Moreover, any of the features described above, such as protrusion, shielding, and/or heat sink features, can be implemented in other disposable sensor designs that are not depicted herein.

The sensors **1600** include an adult/pediatric sensor **1610** for finger placement and a disposable infant/neonate sensor **1602** configured for toe, foot or hand placement. Each sensor **1600** has a tape end **1610** and an opposite connector end **1620** electrically and mechanically interconnected via a flexible coupling **1630**. The tape end **1610** attaches an emitter and detector to a tissue site. Although not shown, the tape end **1610** can also include any of the protrusion, shielding, and/or heat sink features described above. The emitter illuminates the tissue site and the detector generates

a sensor signal responsive to the light after tissue absorption, such as absorption by pulsatile arterial blood flow within the tissue site.

The sensor signal is communicated via the flexible coupling **1630** to the connector end **1620**. The connector end **1620** can mate with a cable (not shown) that communicates the sensor signal to a monitor (not shown), such as any of the cables or monitors shown above with respect to FIGS. **2**A through **2**D. Alternatively, the connector end **1620** can mate directly with the monitor.

FIG. **17** illustrates an exploded view of certain of the components of the sensor **301***f* described above. A heat sink **1751** and a cable **1781** attach to an emitter shell **1704**. The emitter shell attaches to a flap housing **1707**. The flap housing **1707** includes a receptacle **1709** to receive a cylindrical housing **1480**/**1580** (not shown) attached to an emitter submount **1702**, which is attached to a circuit board **1719**.

A spring **1787** attaches to a detector shell **1706** via pins **1783**, **1785**, which hold the emitter and detector shells **1704**, **1706** together. A support structure **1791** attaches to the detector shell **1706**, which provides support for a shielding enclosure **1790**. A noise shield **1713** attaches to the shielding enclosure **1790**. A detector submount **1700** is disposed inside the shielding enclosure **1790**. A finger bed **1710** provides a surface for placement of the patient's finger. Finger bed **1710** may comprise a gripping surface or gripping features, which may assist in placing and stabilizing a patient's finger in the sensor. A partially cylindrical protrusion **1705** may also be disposed in the finger bed **1710**. As shown, finger bed **1710** attaches to the noise shield **1703**. The noise shield **1703** may be configured to reduce noise, such as from ambient light and electromagnetic noise. For example, the noise shield **1703** may be constructed from materials having an opaque color, such as black or a dark blue, to prevent light piping.

Noise shield **1703** may also comprise a thermistor **1712**. The thermistor **1712** may be helpful in measuring the temperature of a patient's finger. For example, the thermistor **1712** may be useful in detecting when the patient's finger is reaching an unsafe temperature that is too hot or too cold. In addition, the temperature of the patient's finger may be useful in indicating to the sensor the presence of low perfusion as the temperature drops. In addition, the thermistor **1712** may be useful in detecting a shift in the characteristics of the water spectrum in the patient's finger, which can be temperature dependent.

Moreover, a flex circuit cover **1706** attaches to the pins **1783**, **1785**. Although not shown, a flex circuit can also be provided that connects the circuit board **1719** with the submount **1700** (or a circuit board to which the submount **1700** is connected). A flex circuit protector **1760** may be provided to provide a barrier or shield to the flex circuit (not shown). In particular, the flex circuit protector **1760** may also prevent any electrostatic discharge to or from the flex circuit. The flex circuit protector **1760** may be constructed from well known materials, such as a plastic or rubber materials.

FIG. **18** shows the results obtained by an exemplary sensor **101** of the present disclosure that was configured for measuring glucose. This sensor **101** was tested using a pure water ex-vivo sample. In particular, ten samples were prepared that ranged from 0-55 mg/dL. Two samples were used as a training set and eight samples were then used as a test population. As shown, embodiments of the sensor **101** were able to obtain at least a standard deviation of 13 mg/dL in the training set and 11 mg/dL in the test population.

FIG. **19** shows the results obtained by an exemplary sensor **101** of the present disclosure that was configured for measuring glucose. This sensor **101** was tested using a turbid ex-vivo sample. In particular, 25 samples of water/glucose/Lyposin were prepared that ranged from 0-55 mg/dL. Five samples were used as a training set and 20 samples were then used as a test population. As shown, embodiments of sensor **101** were able to obtain at least a standard deviation of 37 mg/dL in the training set and 32 mg/dL in the test population.

FIGS. **20** through **22** shows other results that can be obtained by an embodiment of system **100**. In FIG. **20**, 150 blood samples from two diabetic adult volunteers were collected over a 10-day period. Invasive measurements were taken with a YSI glucometer to serve as a reference measurement. Noninvasive measurements were then taken with an embodiment of system **100** that comprised four LEDs and four independent detector streams. As shown, the system **100** obtained a correlation of about 85% and Arms of about 31 mg/dL.

In FIG. **21**, 34 blood samples were taken from a diabetic adult volunteer collected over a 2-day period. Invasive measurements were also taken with a glucometer for comparison. Noninvasive measurements were then taken with an embodiment of system **100** that comprised four LEDs in emitter **104** and four independent detector streams from detectors **106**. As shown, the system **100** was able to attain a correlation of about 90% and Arms of about 22 mg/dL.

The results shown in FIG. **22** relate to total hemoglobin testing with an exemplary sensor **101** of the present disclosure. In particular, 47 blood samples were collected from nine adult volunteers. Invasive measurements were then taken with a CO-oximeter for comparison. Noninvasive measurements were taken with an embodiment of system **100** that comprised four LEDs in emitter **104** and four independent detector channels from detectors **106**. Measurements were averaged over 1 minute. As shown, the testing resulted in a correlation of about 93% and Arms of about 0.8 mg/dL.

Conditional language used herein, such as, among others, "can," "could," "might," "may," "e.g.," and the like, unless specifically stated otherwise, or otherwise understood within the context as used, is generally intended to convey that certain embodiments include, while other embodiments do not include, certain features, elements and/or states. Thus, such conditional language is not generally intended to imply that features, elements and/or states are in any way required for one or more embodiments or that one or more embodiments necessarily include logic for deciding, with or without author input or prompting, whether these features, elements and/or states are included or are to be performed in any particular embodiment.

While certain embodiments of the inventions disclosed herein have been described, these embodiments have been presented by way of example only, and are not intended to limit the scope of the inventions disclosed herein. Indeed, the novel methods and systems described herein can be embodied in a variety of other forms; furthermore, various omissions, substitutions and changes in the form of the methods and systems described herein can be made without departing from the spirit of the inventions disclosed herein. The claims and their equivalents are intended to cover such forms or modifications as would fall within the scope and spirit of certain of the inventions disclosed herein.

What is claimed is:

1. A noninvasive optical physiological measurement device adapted to be worn by a wearer, the noninvasive

optical physiological measurement device providing an indication of a physiological parameter of the wearer comprising:

a plurality of emitters of different wavelengths;

a housing having a surface and a circular wall protruding from the surface;

at least four detectors arranged on the surface and spaced apart from each other, the at least four detectors configured to output one or more signals responsive to light from the one or more light emitters attenuated by body tissue, the one or more signals indicative of a physiological parameter of the wearer; and

a light permeable cover arranged above at least a portion of the housing, the light permeable cover comprising a protrusion arranged to cover the at least four detectors.

2. The noninvasive optical physiological measurement device of claim **1**, wherein the light permeable cover is attached to the housing and forms an airtight or substantially airtight seal enclosing the at least four detectors.

3. The noninvasive optical physiological measurement device of claim **2**, wherein the circular wall creates a gap between the surface and the light permeable cove.

4. The noninvasive optical physiological measurement device of claim **2**, wherein the housing provides noise shielding for the at least four detectors.

5. The noninvasive optical physiological measurement device of claim **4**, wherein the light permeable cover comprises a conductive layer configured to shield the at least four detectors from noise.

6. The noninvasive optical physiological measurement device of claim **3**, wherein the protrusion comprises a continuous protrusion.

7. The noninvasive optical physiological measurement device of claim **6**, wherein the continuous protrusion comprises a convex protrusion.

8. The noninvasive optical physiological measurement device of claim **6**, wherein the light permeable cover is comprised of a rigid material.

9. The noninvasive optical physiological measurement device of claim **8**, wherein the light permeable cover is configured to be positioned between the at least four detectors and tissue of a user when the noninvasive optical physiological measurement device is worn by the user.

10. The noninvasive optical physiological measurement device of claim **9**, wherein the light permeable cover is configured to press against and at least partially deform tissue of the user when the noninvasive optical physiological measurement device is worn by the user.

11. The noninvasive optical physiological measurement device of claim **10**, wherein the light permeable cover is configured to act as a tissue shaper and conform tissue of the user to at least a portion of an external surface shape of the light permeable cover when the noninvasive optical physiological measurement device is worn by the user.

12. The noninvasive optical physiological measurement device of claim **11**, wherein the light permeable cover is configured to reduce a mean path length of light traveling to the at least four detectors.

13. The noninvasive optical physiological measurement device of claim **11**, wherein the at least four detectors are evenly spaced from one another.

14. The noninvasive optical physiological measurement device of claim **1**, wherein the light permeable cover is configured to reduce a mean path length of light traveling to the at least four detectors.

15. The noninvasive optical physiological measurement device of claim **13**, wherein the light permeable cover concentration window is configured to increase a signal to noise ratio of the noninvasive optical physiological measurement device.

16. The noninvasive optical physiological measurement device of claim **13**, wherein the light permeable cover is configured to increase a signal strength per area of the at least four detectors.

17. The noninvasive optical physiological measurement device of claim **1**, wherein the physiological parameter is pulse rate.

18. The noninvasive optical physiological measurement device of claim **1**, wherein the physiological parameter is at least one of: glucose, oxygen, oxygen saturation, methemoglobin, total hemoglobin, carboxyhemoglobin, or carbon monoxide.

19. The noninvasive optical physiological measurement device of claim **1**, wherein the noninvasive optical physiological measurement device is a disposable or a reusable device.

20. The noninvasive optical physiological measurement device of claim **1**, wherein a first detector is arranged spaced apart from a second detector, and a third detector arranged spaced apart from a fourth detector.

21. The noninvasive optical physiological measurement device of claim **20**, wherein the first detector is arranged across a central axis from the second detector and the third detector is arranged across the central axis from the fourth detector, wherein the first, second, third and fourth detectors form a cross pattern about the central axis.

22. The noninvasive optical physiological measurement device of claim **20**, wherein the noninvasive optical physiological measurement device provides a variation in optical path length to the at least four detectors.

23. The noninvasive optical physiological measurement device of claim **1**, wherein the noninvasive optical physiological measurement device is comprised as part of a mobile monitoring device.

24. The noninvasive optical physiological measurement device of claim **23**, wherein the mobile monitoring device includes a touch-screen display.

25. A physiological monitoring system comprising:

the noninvasive optical physiological measurement device of claim **1**; and

a processor configured to receive the one or more signals and communicate physiological measurement information to a mobile phone.

26. A noninvasive optical physiological measurement device adapted to be worn by a wearer providing an indication of a physiological parameter of the wearer comprising:

a plurality of emitters of different wavelengths;

a circular housing comprising a surface with a raised edge;

at least four detectors arranged on the surface, wherein a first detector is arranged spaced apart from a second detector, and a third detector arranged spaced apart from a fourth detector; and

a cover of the circular housing comprising a lens portion, the lens portion comprising a protrusion in optical communication with the at least four detectors,

wherein the at least four detectors are configured to output one or more signals responsive to light from the one or more light emitters attenuated by body tissue, the one or more signals indicative of a physiological parameter of the wearer.

27. The noninvasive optical physiological measurement device of claim **26**, wherein the first detector is arranged

across a central axis from the second detector and the third detector is arranged across the central axis from the fourth detector, wherein the first, second, third and fourth detectors form a cross pattern about the central axis.

28. The noninvasive optical physiological measurement device of claim **26**, wherein the at least four detectors are arranged in a grid pattern such that the first detector and the second detector are arranged across from each other on opposite sides of a central point along a first axis, and the third detector and the fourth detector are arranged across from each other on opposite sides of the central point along a second axis which is perpendicular to the first axis.

29. The noninvasive optical physiological measurement device of claim **27**, wherein the physiological parameter is pulse rate.

30. The noninvasive optical physiological measurement device of claim **27**, wherein the physiological parameter is at least one of: glucose, oxygen, oxygen saturation, methemoglobin, total hemoglobin, carboxyhemoglobin, or carbon monoxide.

* * * * *

UNITED STATES PATENT AND TRADEMARK OFFICE

CERTIFICATE OF CORRECTION

PATENT NO. : 10,258,265 B1
APPLICATION NO. : 16/212440
DATED : April 16, 2019
INVENTOR(S) : Jeroen Poeze et al.

It is certified that error appears in the above-identified patent and that said Letters Patent is hereby corrected as shown below:

In the Specification

In Column 25 at Line 18, After "where" insert --R--.

In Column 40 at Line 39 (approx.), After "*Noise* = $\sqrt{aR + bR^2}$, " insert --where:--.

In Column 41 at Line 35, Change "MO" to --MΩ--.

In Column 41 at Line 38, Change "MO" to --MΩ--.

In Column 41 at Line 47, Change "MO" to --MΩ--.

In Column 41 at Line 49, Change "MO" to --MΩ--.

In the Claims

In Column 45 at Line 22, In Claim 3, change "cove." to --cover.--.

In Column 45 at Line 63, In Claim 14, change "claim 1," to --claim 13,--.

In Column 46 at Line 1, In Claim 15, before "is" delete "concentration window".

Signed and Sealed this
Twenty-second Day of October, 2019

Andrei Iancu
Director of the United States Patent and Trademark Office