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(54) **WHOLE BLOOD ASSAY**

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(57) **ABSTRACT**

A method and apparatus to estimate the concentration of a target substance (e.g. Cholesterol or CRP) in the plasma component of a whole blood sample without the need to separate the red blood cells from the plasma prior to testing, thereby simplifying the design and construction of the test device. The invention achieves this by measuring the analyte under investigation in a time dependent (bio-/immuno-) chemical reaction and measuring separately, a marker substance (e.g haemoglobin) for the estimation of red blood cell volume, using a non-time-dependent alteration in physical property of the reaction mixture (in this instance, transmission) attributed to inherent filter effects on sample addition. These non-time- dependent changes are not part of the reaction chemistry, and are resolved from the time dependent alteration in physical property caused by the assay chemistry by continuous measurement and mathematical modelling. Algorithms that combine these two parameters are used to estimate the target substance and compensate for variations in the percentage haematocrit of the sample. The method equalises the assay response for subtle variations in patient sample (e.g. haematocrit).

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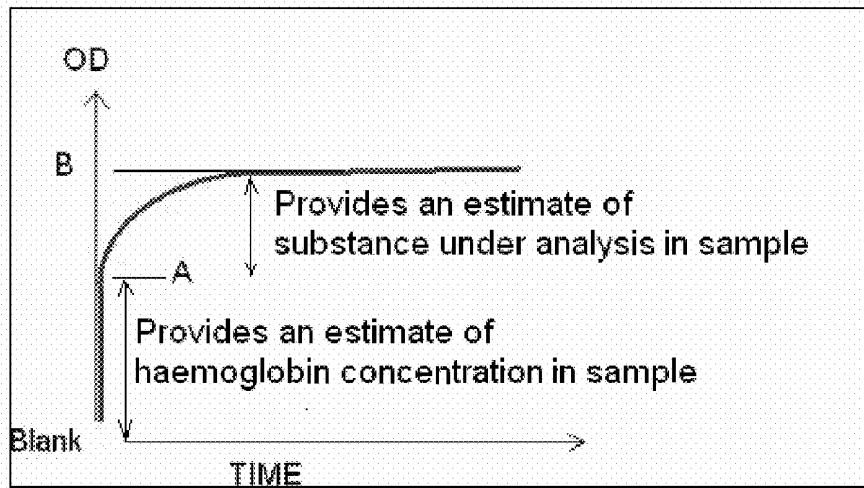
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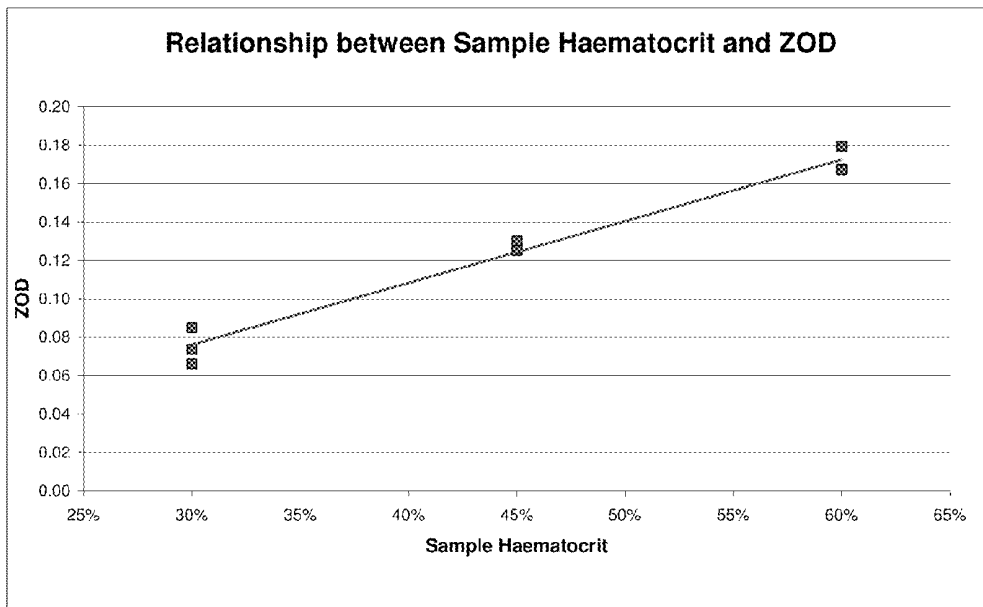
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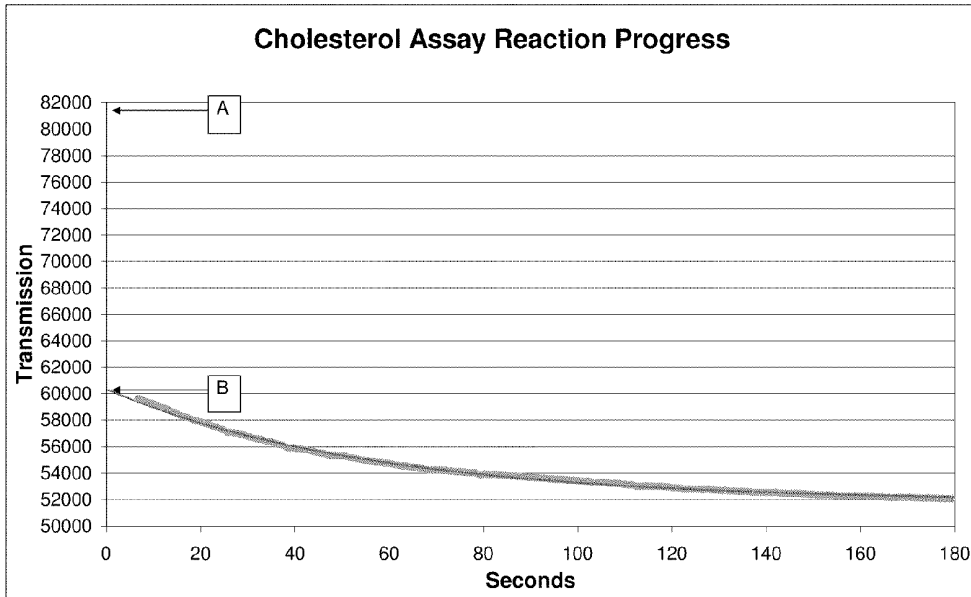
**Figure 1**



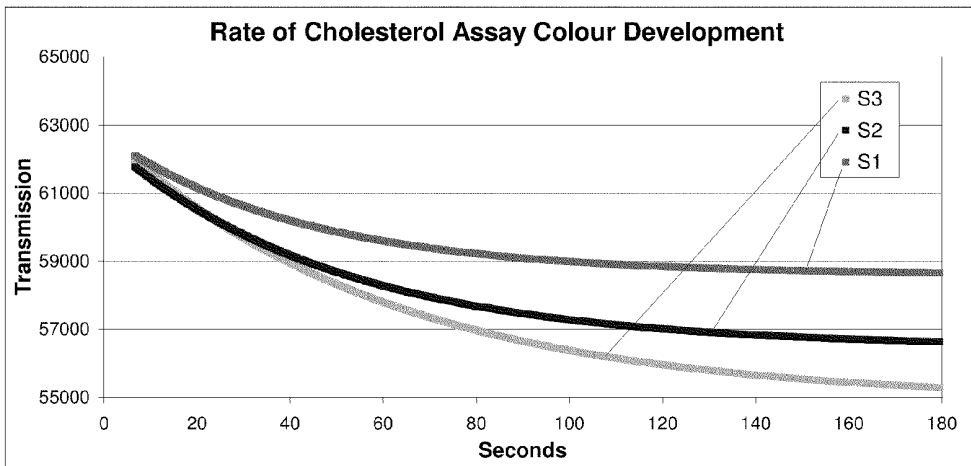
**Figure 2**



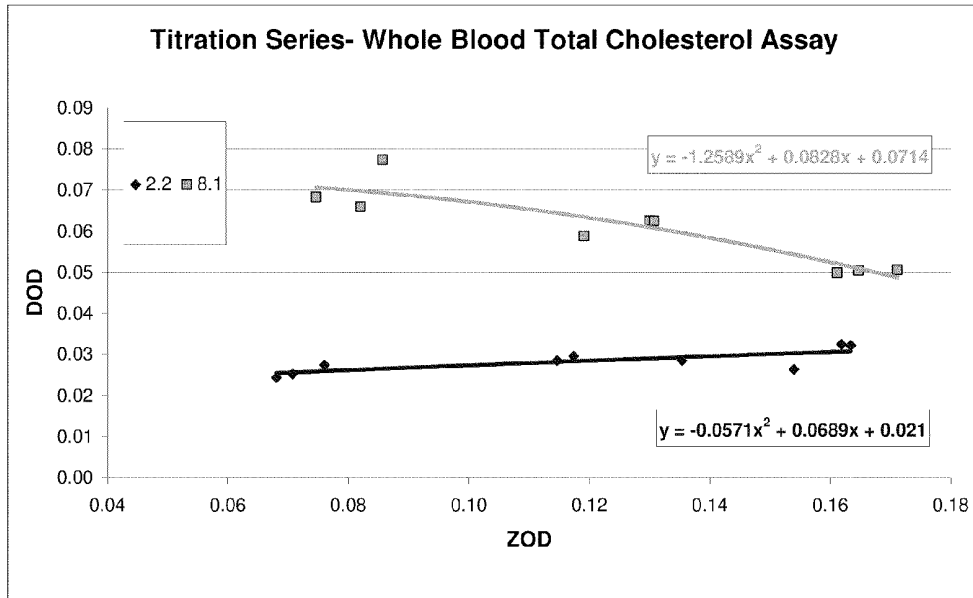
**Figure 3**



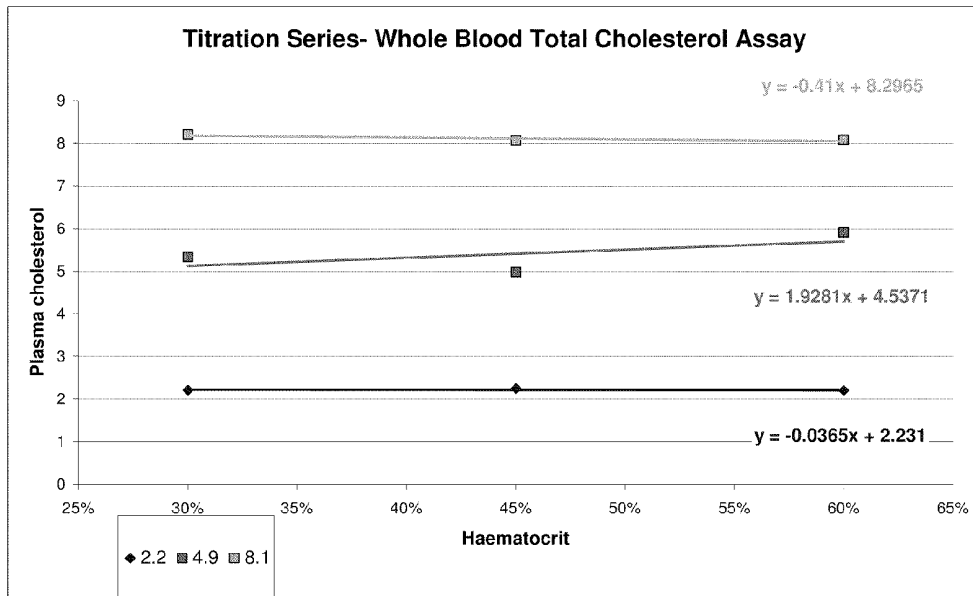
**Figure 4**



**Figure 5**



**Figure 6**



## WHOLE BLOOD ASSAY

**[0001]** The present invention relates to methods of carrying out an assay for a substance in a whole blood sample.

**[0002]** UK patent application number 0606450.5 describes the assay of glycosylated proteins by fluorescence quenching using a method and apparatus to estimate the concentration of a non-fluorescent substance (e.g. haemoglobin) in a fluorescent assay by separately estimating the non-time-dependent alteration attributed to inherent filter effects from the time dependent alteration caused by the assay chemistry (measuring haemoglobin-A1c). Such a method obviates the requirement for a separate photometric or other measurement thereby simplifying the methodology and associated instrumentation.

**[0003]** The present invention is an adaptation of this approach for the measurement of other substances, not necessarily by fluorimetry, whereby the developing signal (e.g. transmission) is monitored continuously and used, not only to measure the substance undergoing reaction, but also to measure and compensate for the initial amount of sample added.

**[0004]** The prior art involves the assay of a range of substances whereby a specific reaction chemistry is followed photometrically with time, for example by utilising an antibody specific to an analyte that is coated onto microfine latex particles and measuring the increased turbidity that is produced when the analyte being measured promotes aggregation of the latex particles as the reaction between analyte/antigen and antibody proceeds. This measurement of increasing turbidity can be achieved using a conventional photometer and using the associated scientific principles of photometric measurements. Such concentration dependent turbidity is then compared to that produced by standards which is established prior art.

**[0005]** Further compatible methodologies include carrying out a series of enzyme-linked reactions in solution, where an analyte in the plasma fraction of a whole blood sample is altered by an enzyme-promoted reaction to ultimately derive a coloured dye from colourless reaction constituents. The colour is developed in a time dependent way and monitored photometrically. This measurement of colour change can also be achieved using a conventional photometer using the associated scientific principles of photometric measurements. Such concentration dependent change in transmission is then compared to that produced by standards which method is also established prior art.

**[0006]** In whole blood samples there is a variable which must be taken into account when analysing substances that are present in the plasma component. This is the haematocrit or percentage of red blood cells by volume in the whole blood sample and this value can vary widely depending on age, climate, nutritional and disease status and other factors. For example, a 40% haematocrit means that in a given volume of whole blood, 40% of that volume is attributed to the volume taken up by red blood cells and 60% to plasma. As the haematocrit of a patient's blood rises, so the volume of plasma in a fixed volume sample which is introduced into the test device decreases and vice versa. Since it is the plasma component which exclusively carries the analyte being measured, then the lower the volume of plasma component added to the reaction mix, the lower the resulting concentration of the substance being measured in that reaction mix and the resulting assayed value and vice versa.

**[0007]** Any analysis that produces a concentration of a plasma substance in whole blood must be corrected for variations in haematocrit to give a true plasma concentration. It is for this reason that such assays are conventionally performed on serum or plasma that has been separated previously from red blood cells by filtration or centrifugation. In a point of care, doctor's office or clinic setting, small volumes of whole blood sample may be separated from the plasma component by filtration or other mechanical manipulations which add complexity, and hence cost, into the design of the system.

**[0008]** It would be most useful in these situations to measure two substances, one of which is the analyte under investigation and the other which is considered to be a marker by which to estimate or normalise the sample haematocrit.

**[0009]** It is known that the haemoglobin concentration of whole blood, after red blood cells are lysed, is directly proportional to the red blood cell volume in the whole blood sample.

**[0010]** Haemoglobin concentration can be estimated photometrically by known means at various wavelengths or by various reaction chemistries as described above (i.e. turbidity or enzyme-catalysed colour formation). An instrument that measures transmission of a sample continuously, after allowing for the necessary blanking measurements initially, at a point in the visible spectrum where both assay chemistry (either colour formation or development of turbidity) and haemoglobin can be measured may, by the use of the algorithms described in UK patent application 0606450.5, make estimates of the initial transmission, and hence derive an estimate of haemoglobin concentration (since this effect is instantaneous and not dependent on any chemical reaction) before the analyte reaction chemistry progresses. The same algorithm also makes a measurement of the final transmission which would, by differential analysis, represent the time dependent chemistry undergone by the analyte under investigation. An algorithm that computes the relationship between these two estimates may be used to produce a plasma value that is not influenced by variations in haematocrit.

**[0011]** According to a first aspect of the present invention there is provided a method of carrying out an assay for substance in a sample of whole blood, which method comprises:

**[0012]** (a) carrying out a reaction in solution between an analyte in the whole blood sample and a specific reagent, the sample being combined with the reagent at time  $t_0$ ;

**[0013]** (b) monitoring the transmission of the solution continuously as the reaction progresses at a suitable wavelength;

**[0014]** (c) recording, from the detected transmission values, the transmission,  $T_{init}$ , being the transmission of the reaction solution before the addition of the sample; and calculating  $T_0$ , being the transmission at time  $t_0$  after the sample has been added and  $T_\infty$  being the transmission at time  $t_\infty$ , which is the point at which all of the analyte has reacted with the reagent, or has attained equilibrium;

**[0015]** (d) calculating from the values of  $T_{init}$  and  $T_0$  the optical density of the sample after blood addition and hence quantity of haemoglobin added in that sample;

**[0016]** (e) calculating from the values of  $T_0$  and  $T_\infty$  the change in transmission attributable to the reaction in step (a); and

**[0017]** (f) deriving from the relationship between these measurements the concentration of the analyte in the sample, corrected for haematocrit variation.

**[0018]** A currently intended major use of the present invention is in the assaying of plasma analytes in whole blood. Two

examples are described below to illustrate the principles applied to the measurement of an analyte such as C-reactive Protein (CRP) by an immunoturbidimetric method, or to a different analyte such as cholesterol by a series of enzyme driven linked reactions to produce a coloured end point.

**[0019]** In the immunoturbidimetric example, preferably the antibody to CRP is bound to a particle, such as a latex bead, that will aggregate as the reaction between antibody and analyte progresses.

**[0020]** In the enzyme-driven series of linked reactions, preferably the key enzymes are specific to the substance being measured such as cholesterol esterase and cholesterol oxidase.

**[0021]** The relationship derived from the measurements of  $T_0$  and  $T_\infty$  referred to in step (f) will preferably be achieved using an algorithm of the form:

$$y = ax^2 + bx + c$$

**[0022]** where a, b and c are calibration constants, y is a measure of analyte chemistry e.g.  $\log [(T_0 - T_\infty) / T_0]$  and x is a measure of haemoglobin e.g.  $\log [(T_{init} - T_0) / T_{init}]$ .

**[0023]** In a fully automated instrument system, where blood is added to the reagent and mixed by the instrument, the earliest transmission point detectable as the reaction chemistry progresses might possibly be determined within the first 5 seconds or so after blood addition and mixing. However, in a non-automated system, the monitoring of the initial binding reaction is further delayed due to the finite time required by an operator physically to add the blood specimen to the reaction cuvette containing the reagent, then mix and return the reaction cuvette to the photometer. Therefore, the actual transmission level at  $t_0$ , immediately after the addition of the blood specimen, cannot directly be measured in either the manual or automatic system.

**[0024]** To overcome this, the transmission level  $T_0$  at time  $t_0$  i.e. immediately after sample addition and mixing but before reaction has occurred, is preferably determined by back extrapolation of a curve fitted to the time course transmission data, based upon the rate equation of the chemical binding reaction. This, preferably, includes plotting the detected transmission data against time, and applying a best fit curve to the plotted points. The plotting may be physical or even manual, but is often achieved automatically and mathematically by fitting a mathematical function of a curve of best fit to the data and extrapolating that function to time  $t_0$  and  $t_\infty$ , without the generation of a graph as such. The curve fitting may be achieved by any suitable mathematical method, and an example method is described later in this specification. Using these regimes the recorded data is used to minimise variance and then the curve is extrapolated to provide values for  $T_0$ .

**[0025]** Further, by forward extrapolation of this curve fit beyond the data collection period, the transmission level  $T_\infty$  at time  $t_\infty$ , the reaction end-point, is also determined. Determining the transmission levels at  $t_0$  and  $t_\infty$  in this manner has been found to produce reliable and accurate results.

**[0026]** The time period over which the transmission of the reaction mixture is measured may be anything suitable for the use to which the assay is put. A long period gives greater accuracy, but a slower assay process; whereas a short period is convenient, but may lead to less precise results as the amount of data from which to extrapolate becomes sparse. The suitable length of the measurement period will depend on the substance being assayed and the time profile of its reaction

with the antibody. In respect of both the CRP immunoturbidimetric assay and the enzymatic cholesterol assay using readily available reagents, a measurement period of about 3 minutes is usually appropriate. It is also possible to shorten the time period by extrapolating the results from less data and continually checking the accuracy of the extrapolation as the reaction progresses. If the data from early extrapolation (i.e. a short period such as 20-30 seconds) is being substantiated by the continuing recorded observation, the original extrapolation may be used to reliably predict the end point and start point result from less data and in a shorter period of time.

**[0027]** It is preferred that before the sample is combined with the antibody reagent in step (a), the solution alone is excited by incident electromagnetic radiation of a suitable wavelength  $\lambda_n$  (i.e. a wavelength where transmission measurements can detect both haemoglobin and either the turbidity produced by the antibody-analyte reaction or the colour produced by the enzyme driven linked reactions) and the resultant initial transmission ( $T_{init}$ ) at the selected wavelength frequency is detected. This in combination with the value of  $T_0$  can be used to calculate the Zero Optical Density (ZOD) using the equation:

$$ZOD = \text{Log} [T_{init} / T_0]$$

**[0028]** The ZOD is known to be directly proportional to the total haemoglobin concentration. The measurement of initial transmission can therefore be used to determine total haemoglobin concentration, which, in turn, may be used to determine red blood cell volume.

**[0029]** It may not be necessary to determine  $T_{init}$  before every assay, as this value may be a standard or constant value depending on the reproducibility of the reaction cuvette and the reagents used.

**[0030]** Many modifications of the methods and apparatus of the present invention described herein will be apparent to the skilled man and these fall within the scope of the present invention. Fundamental to the invention however is that the physical parameter being measured (be it fluorescence, transmission or some other physical property such as turbidity) is altered in a time dependent fashion by the specific reaction of one substance on the one hand and in an instantaneous, and therefore non-time dependent, fashion by another substance on the other; one substance being the target analyte being measured, another substance being the marker by which variations in sample addition or property are compensated for.

**[0031]** However in order that its principles may be better understood, but by way of example only, the present invention will now be described in detail with reference to the example of measuring cholesterol in whole blood and where appropriate to the accompanying drawings in which:

**[0032]** FIG. 1 is a graph of a general profile of a reaction in which Optical Density (OD) increases due to the specific reaction chemistry as reaction time progresses after an initial increase in OD (or drop in transmission) attributed to the presence of the sample itself after initial blank measurement;

**[0033]** FIG. 2 is a graph depicting the relationship between Zero Optical Density (ZOD) at 510 nm derived from the initial drop in transmission (i.e. at time zero) and quantity of haemoglobin (from a fixed volume of sample at varying haematocrit levels) added;

**[0034]** FIG. 3 is a plot of the transmission signal time course during the development of a coloured response in

solution from cholesterol in the sample and a series of enzyme-linked reactions that alter a colourless precursor into a coloured dye;

[0035] FIG. 4 is a composite plot of the time course of reduction in transmission during the reaction between various levels of cholesterol in solution and a series of cholesterol-specific, enzyme-linked reactions in the presence of blood;

[0036] FIG. 5 shows the relationship between the Delta Optical Density (DOD), being the difference between  $OD_0$  and  $OD_\infty$  of the reaction chemistry, and Zero Optical Density (ZOD), being the measure of haemoglobin (and hence haematocrit), for a low (2.2 mmol/litre) and high (8.1 mmol/litre) standard blood sample; and

[0037] FIG. 6 shows the cholesterol values derived from the algorithm and reaction chemistry method as they vary with sample haematocrit for three standards covering the assay working range.

[0038] The present invention may be used to assay cholesterol in blood. In such a method the constituents of a series of enzyme-linked reactions dissolved in a suitable buffer are introduced into a cuvette. Prior to the introduction of a blood specimen, this reagent mixture is excited in a photometer by EM radiation at a suitable wavelength (510 nm) and the transmission blank ( $T_{mit}$ ) is measured.

[0039] The cuvette is removed from the photometer or left in place and the blood sample is immediately added and mixed and the transmission over a time course is detected and recorded. This data is plotted and a curve is fitted to the data set. FIG. 3 shows such a reaction time profile where the initial transmission  $T_{mit}$  labelled A is recorded prior to sample introduction at zero seconds. The actual experimental transmission data for the reaction is recorded over time (usually at less than 1 second intervals) from the reintroduction of the cuvette in a manual system or after mixing in an automatic system until a suitable period has elapsed. By back extrapolation, the transmission level  $T_0$  is determined at  $t_0$ , i.e. the point when the sample was added but no reaction with the target analyte had occurred (B). By forward extrapolation (which is not shown because the end of the measured data is off the graph in FIG. 3) the transmission level  $T_\infty$  at the reaction end-point  $t_\infty$  is similarly determined wherein the reaction with the target analyte has attained completion.

[0040] Of many potential curve fit routines that have been shown to be effective in the extrapolated estimation of  $T_0$  and  $T_\infty$  a suitable curve fitting routine is based upon the general rate equation below:

$$T_t = T_0 + (T_\infty - T_0) \times (1 - e^{-t/\theta})$$

[0041] where,

[0042]  $T_t$  = Transmission at time t seconds

[0043]  $T_0$  = Transmission at time zero

[0044]  $T_\infty$  = Transmission at time infinity (i.e. at reaction end)

[0045]  $e$  = 2.7183 (natural log base)

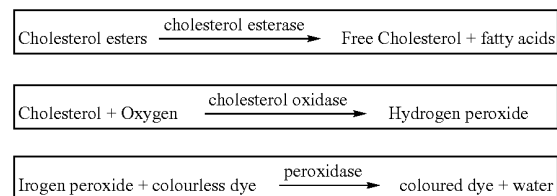
[0046]  $\theta$  = rate constant

[0047] with  $T_0$ ,  $T_\infty$  and  $\theta$  being determined iteratively by minimising the sum of the squared variances between the fitted and measured value at each one second data point. i.e.  $\sum (T_{t-actual} - T_{t-estimated})^2$  is minimised by the fitting routine.

[0048] The mathematical modelling of the data can be achieved by other methods of curve fitting which may be equally acceptable in practice.

#### Example

[0049] A known volume (e.g. 2.0 ml) of reagent comprising a buffer and a mixture of enzyme and reactants for the enzyme-linked reactions (see below) was introduced into a reaction cuvette.



[0050] This mixture was introduced into a photometer and was excited at 510 nm. Blank transmission readings were taken at <1 second intervals for about five seconds.

[0051] A fixed volume (5  $\mu$ L) of blood, whose plasma cholesterol level was known, was then introduced into the cuvette at time  $t_0$  and mixed. When the mixing ceased the transmission at 510 nm was then measured for 3 minutes reaction time. The time taken to effect mixing and for the vortex in the liquid to stop means that the transmission was only recorded from about the 10th second after the addition of the blood sample onwards.

[0052] An example of the blank and blood readings of transmission signal and reference are laid out in Table 1 below.

TABLE 1

Reading time	Transmission
0.42	81679
0.79	81715
1.16	81729
1.52	81736
1.89	81731
2.26	81740
2.63	81722
2.99	81734
3.36	81729
3.73	81761
4.1	81776
4.47	81785
4.83	81779
5.2	81773

TABLE 1-continued

Column 1 Time after blood addition (secs)	Column 2 T	Column 3 Mathematically fitted curve	Column 4 Time after blood addition (secs)	Column 5 T	Column 6 Mathematically fitted curve	Column 7 Time after blood addition (secs)	Column 8 T	Column 9 Mathematically fitted curve
0.42		56090	60.36	49933	50023	120.29	47761	47798
1.16		55972	61.09	49901	49980	121.03	47713	47782
2.26		55799	62.2	49905	49916	122.13	47614	47759
3.36		55630	63.3	49823	49854	123.24	47600	47736
4.47		55462	64.4	49757	49793	124.34	47618	47714
5.57		55299	65.51	49683	49732	125.44	47574	47692
6.67		55138	66.61	49643	49674	126.55	47575	47670
7.41		55032	67.34	49598	49635	127.28	47552	47656
8.14		54928	68.08	49594	49597	128.02	47510	47642
9.25		54774	69.18	49525	49540	129.12	47530	47621
10.35	54801	54623	70.29	49469	49485	130.22	47520	47601
11.08	54610	54524	71.02	49452	49449	130.96	47566	47587
12.19	54594	54377	72.12	49427	49395	132.06	47652	47568
13.29	54443	54233	73.23	49413	49342	133.17	47690	47548
14.06	54352	54134	73.96	49372	49307	133.9	47755	47536
15.13	54200	53999	75.07	49296	49256	135	47781	47517
16.23	53996	53862	76.17	49296	49206	136.11	47774	47498
17.34	53869	53727	77.27	49223	49157	137.21	47754	47480
18.44	53678	53595	78.38	49181	49108	138.31	47729	47463
19.91	53449	53423	79.85	49134	49045	139.78	47688	47440
20.64	53331	53339	80.58	49087	49014	140.52	47694	47428
21.75	53242	53213	81.69	49017	48968	141.62	47690	47411
22.85	53125	53091	82.79	49028	48923	142.73	47676	47395
23.95	52990	52971	83.89	48979	48879	143.83	47608	47379
25.06	52862	52852	85.01	48916	48835	144.93	47539	47363
26.16	52736	52736	86.1	48938	48793	146.04	47457	47347
27.29	52620	52620	87.2	48945	48751	147.14	47258	47332
28	52509	52548	87.94	48871	48724	147.87	47088	47322
29.1	52405	52437	89.04	48896	48683	148.98	47030	47307
30.21	52275	52328	90.14	48874	48644	150.08	47152	47292
30.57	52252	52293	90.51	48867	48631	150.45	47218	47287
31.31	52170	52222	91.25	48835	48604	151.18	47278	47278
32.78	52016	52083	92.72	48777	48554	152.65	47313	47259
33.15	51954	52049	93.08	48775	48541	153.02	47277	47255
34.99	51684	51881	94.92	48731	48480	154.86	46858	47232
35.35	51650	51849	95.29	48740	48468	155.23	46763	47228
36.82	51545	51719	96.76	48708	48420	156.7	46498	47210
37.93	51428	51623	97.86	48600	48385	157.8	46318	47197
38.3	51405	51592	98.24	48586	48373	158.17	46304	47193
39.77	51196	51468	99.7	48508	48328	159.64	46324	47177
40.13	51140	51438	100.07	48493	48317	160.01	46370	47172
41.97	51052	51289	101.91	48376	48262	161.85	46584	47152
42.34	51056	51259	102.28	48326	48251	162.21	46638	47149
43.81	50925	51144	103.75	48292	48209	163.69	46544	47133
44.18	50903	51115	104.12	48260	48199	164.05	46487	47129
45.65	50775	51003	105.59	48302	48157	165.52	46519	47114
46.02	50673	50976	105.95	48304	48148	165.89	46544	47110
47.86	50253	50840	107.79	48278	48098	167.73	46587	47092
48.22	50203	50814	108.16	48281	48088	168.1	46549	47089
49.69	50152	50709	109.63	48247	48050	169.57	46668	47074
50.06	50096	50684	110	48268	48040	169.94	46677	47071
51.9	49943	50557	111.84	48246	47994	171.77	46680	47054
52.27	49926	50532	112.21	48230	47985	172.14	46674	47051
53.74	50013	50434	113.68	48199	47949	173.61	46724	47038
54.11	50035	50410	114.04	48193	47940	173.98	46768	47034
55.95	50102	50292	115.88	48074	47897	175.82	46703	47018
56.31	50115	50269	116.28	48036	47888	176.19	46694	47015
57.78	50095	50178	117.72	47976	47855	177.66	46730	47003
58.15	50087	50155	118.09	47967	47846	178.03	46751	47000
59.99	49959	50044	119.93	47794	47806	179.86	46780	46985
60.36	49933	50023	120.29	47761	47798	180.23	46774	46982

**[0053]** Columns 2, 4, 6 and 8 are the measured transmission readings. If this data is plotted against time it produces a curve as shown in FIG. 3. A mathematically derived curve is fitted to these values using a suitable algorithm as previously discussed and the values (as shown in columns 3, 5, 7 and 9) are also plotted on the graph in FIG. 3. The values  $T_0$  and  $T_\infty$  can

be derived for this sample by back and forward extrapolation of the data fit to  $t_0$  and  $t_\infty$  respectively.

**[0054]** Table 2 shows the results of a series of equivalent assays run on blood samples where the haematocrit has been deliberately manipulated to cover the normal range of 30% to 60%.

TABLE 2

Column 1 Cholesterol Target	Column 2 Haematocrit	Column 3 ZOD	Column 4 DOD	Column 5 Cholesterol calculated	Column 6 Mean	Column 7 CV
2.2	30%	0.068	0.024	2.06	2.21	7.91%
2.2	30%	0.076	0.027	2.40		
2.2	30%	0.071	0.025	2.16		
2.2	45%	0.117	0.030	2.41	2.24	7.74%
2.2	45%	0.115	0.029	2.26		
2.2	45%	0.135	0.029	2.06		
2.2	60%	0.163	0.032	2.63	2.20	37.37%
2.2	60%	0.154	0.026	1.25		
2.2	60%	0.162	0.032	2.71		
4.9	30%	0.066	0.049	5.21	5.33	6.83%
4.9	30%	0.074	0.047	5.04		
4.9	30%	0.085	0.052	5.74		
4.9	45%	0.125	0.044	4.93	4.97	0.76%
4.9	45%	0.126	0.044	5.00		
4.9	45%	0.130	0.044	4.99		
4.9	60%	0.167	0.043	5.88	5.91	17.11%
4.9	60%	0.167	0.040	4.91		
4.9	60%	0.179	0.043	6.93		
8.1	30%	0.082	0.066	7.59	8.20	10.78%
8.1	30%	0.086	0.077	9.21		
8.1	30%	0.075	0.068	7.80		
8.1	45%	0.130	0.063	8.42	8.06	7.83%
8.1	45%	0.119	0.059	7.33		
8.1	45%	0.131	0.062	8.43		
8.1	60%	0.171	0.051	8.75	8.08	7.77%
8.1	60%	0.161	0.050	7.50		
8.1	60%	0.165	0.051	7.98		

**[0055]** For each sample, the Zero Optical Density (ZOD) and Delta Optical Density (DOD) are calculated from  $T_{Blank}$ ,  $T_0$  and  $T_\infty$  using the formulae below;

$$ZOD = \text{Log} [(T_{Blank})/(T_0)]$$

$$DOD = [\log (T_0/T_\infty)] - ZOD$$

**[0056]** A series of plots of the relationship between DOD and ZOD as the haematocrit is changed for standards at the lower and upper limit of the assay working range are described mathematically using a second order polynomial curve fit. The calibration constants a, b & c (and a', b' & c') are defined (second order polynomial) for these standards of known high and low plasma cholesterol which have been manipulated to produce different haematocrit levels (see FIG. 5).

**[0057]** The plasma concentration of cholesterol for unknown samples can then be calculated from experimentally derived values of  $DOD_{UNKNOWN}$  and  $ZOD_{UNKNOWN}$  thus;

**[0058]** 1. Calculate DOD for High Standard (H) from ZOD of sample

$$DOD_{HIGH} = a \cdot ZOD^2 + b \cdot ZOD + c$$

**[0059]** 2. Calculate DOD for Low Standard from (L) ZOD of sample

$$DOD_{LOW} = a' \cdot ZOD^2 + b' \cdot ZOD + c'$$

**[0060]** 3. Calculate plasma cholesterol for unknown sample from the following expression:

$$\frac{[DOD_{UNKNOWN} - DOD_{LOW}]/[DOD_{HIGH} - DOD_{LOW}] \times [H - L] + L}{[H - L]}$$

**[0061]** The utility of this approach to equalising cholesterol values for the presence of red blood cells is demonstrated by back calculating the cholesterol values for the high and low

standards and also of a sample which is approximately mid range between high and low standards. (Table 2; columns 5, 6 and 7 and FIG. 6.)

**[0062]** This shows that the derived values are largely unaffected by sample haematocrit.

1. A method of carrying out an assay for substance in a sample of whole blood, which method comprises:

- carrying out a reaction in solution between an analyte in the whole blood sample and a specific reagent, the sample being combined with the reagent at time  $t_0$ ;
- monitoring the transmission of the solution continuously as the reaction progresses at a suitable wavelength;
- recording, from the detected transmission values, the transmission,  $T_{init}$ , being the transmission of the reaction solution before the addition of the sample; and calculating  $T_0$ , being the transmission at time  $t_0$  after the sample has been added and  $T_\infty$  being the transmission at time  $t_\infty$ , which is the point at which all of the analyte has reacted with the reagent, or has attained equilibrium;
- calculating from the values of  $T_{init}$  and  $T_0$  the optical density of the sample after blood addition and hence quantity of haemoglobin added in that sample;
- calculating from the values of  $T_0$  and  $T_\infty$  the change in transmission attributable to the reaction in step (a); and
- deriving from the relationship between these measurements the concentration of the analyte in the sample, corrected for haematocrit variation.

2. A method as claimed in claim 1, wherein the reaction in step (a) is an immunoturbidimetric one.

3. A method as claimed in claim 2 wherein the reagent includes an antibody specific to the analyte bound to a particle.

4. A method as claimed in claim 3 wherein the particle is a latex bead.

5. A method as claimed in claim 1, wherein the analyte is C-reactive Protein.

6. A method as claimed in claim 1, wherein the reaction in step (a) is colourimetric

7. A method as claimed in claim 6, wherein the reagent includes enzymes to produce an enzyme driven series of linked reactions to produce a coloured end point.

8. A method as claimed in claim wherein the analyte is cholesterol.

9. A method as claimed in wherein the reagent includes one or both of cholesterol oxidase and cholesterol esterase.

10. A method as claimed in claim 1 wherein the algorithm in step (f) is of the form  $y=ax^2+bx+c$ , where a, b and c are calibration constants, y is a measure of analyte chemistry from the calculation of  $\log [(T_0-T_\infty)/T_0]$  and x is a measure of haemoglobin from the calculation of  $\log [(T_{mit}-T_0)/T_{mit}]$ .

11. A method as claimed in claim 1, wherein  $T_0$  is determined by back extrapolation of a curve fitted to time course transmission data, based upon the rate equation of the reaction.

12. A method as claimed in claim 1, wherein  $T_\infty$  is determined by forward extrapolation of this curve fit beyond the data collection period.

13. A method as claimed in claim 11, wherein the extrapolation includes plotting the detected transmission data against time, and applying a best fit curve to the plotted points.

14. A method as claimed in claim 13, wherein the plotting is achieved by fitting a mathematical function of a curve of best fit to the data and extrapolating that function to time  $t_0$  and  $t_\infty$ .

15. A method as claimed in claim 1 wherein the period during which transmission data is recorded is up to 3 minutes.

16. A method as claimed in claim 1, wherein before the sample is combined with the reagent in step (a), the sample alone is excited by incident electromagnetic radiation of a wavelength  $\lambda_n$ , wherein the wavelength  $\lambda_n$  can detect both haemoglobin and the reaction and the resultant initial transmission ( $T_{mit}$ ) at the selected wavelength frequency is detected.

17. A method as claimed in claim 16, wherein  $T_{mit}$  in combination with the value of  $T_0$  can be used to calculate the Zero Optical Density (ZOD) using the equation  $ZOD=\text{Log} [(T_{mit}/T_0)]$ .

\* \* \* \* \*

专利名称(译)	全血分析		
公开(公告)号	<a href="#">US20100075338A1</a>	公开(公告)日	2010-03-25
申请号	US12/532596	申请日	2008-03-20
申请(专利权)人(译)	QUOTIENT诊断有限公司		
当前申请(专利权)人(译)	EKF诊GMBH		
[标]发明人	VESSEY JOHN PHILLIP GRAY ADRIAN RICHARD PERCIVAL DAVID		
发明人	VESSEY, JOHN PHILLIP GRAY, ADRIAN RICHARD PERCIVAL, DAVID		
IPC分类号	G01N33/53 C12Q1/02 C12Q1/60		
CPC分类号	G01N33/49 G01N33/72 G01N2333/918 G01N2333/904 G01N33/92		
优先权	2007005495 2007-03-22 GB		
其他公开文献	US7943331		
外部链接	<a href="#">Espacenet</a> <a href="#">USPTO</a>		

摘要(译)

一种估计全血样品血浆成分中目标物质(例如胆固醇或CRP)浓度的方法和装置,无需在测试前将红细胞与血浆分离,从而简化了血浆成分的设计和构造。测试设备。本发明通过在时间依赖性(生物/免疫)化学反应中测量被研究的分析物并单独测量,使用非时间依赖性测量红细胞体积的标记物质(例如血红蛋白)来实现这一点。反应混合物的物理性质的改变(在这种情况下,透射)归因于对样品添加的固有过滤效应。这些非时间依赖性变化不是反应化学的一部分,并且通过连续测量和数学建模从测定化学引起的物理性质的时间依赖性改变中解决。结合这两个参数的算法用于估计目标物质并补偿样品的血细胞比容百分比的变化。该方法使测定响应等于患者样品中的细微变化(例如血细胞比容)。

