Title: RESPIRATION AND HEART RATE MONITOR

Abstract: Respiration rate is a challenging parameter to measure. Traditional techniques tend to be uncomfortable for the patient. Therefore, there is provided a respiration rate monitor (20) for measuring the respiration rate of a body (30), a monitor (20) comprising at least one accelerometer (10) attachable to an upper part of a body (30) and arranged to provide an accelerometer output signal which is related to the respiration rate of that body (30), and means (40) for determining the respiration rate from the or each accelerometer (10) output signal and for providing a monitor output (50) indicative of a respiration rate. The means (40) comprising a baseline filter (110) for removing artefacts of the accelerometer output signal which are not a result of the respiration of the body, so as to provide a removed baseline signal.
Respiration and Heart Rate Monitor

This invention relates to a monitor for determining respiration and/or heart rate, typically in humans.

Respiration rate is a challenging parameter to measure. A number of traditional techniques exist, but each suffers from drawbacks. In one well known technique, a clinician or nurse counts the number of breaths a patient takes in a period of time. Apart from being a time-consuming process, this process is difficult because it must be done without the patient being aware that their respiration is being observed. If they become aware, then their breathing pattern may change either consciously or sub-consciously.

As an alternative, a patient’s airways may be connected to tubes which detect flow rate therethrough. Either a mask is placed over the face of the patient or a tube is inserted down their throat. The use of a mask is unpleasant, particularly when a patient is short of breath, and the insertion of a tube down a patient’s throat is invasive.

One or more elastic belts may be placed around the patient. Sensors in the belt or belts determine the change in chest volume and respiration rate may be determined from this. Again, many patients find the fitting of elastic belts to be unpleasant.

Measuring heart rate is a more straightforward problem and is typically addressed by observing the electrical activity of the heart using electrodes placed on the patient’s chest (electrocardiography or ECG), or by measuring blood oxygenation at, for example, a finger tip (photoplethysmography). Nevertheless, these heart rate measuring devices are not suitable for measurement of breathing rate.

It would accordingly be desirable to provide an improved respiration rate monitor. It would further be
desirable to provide an improved heart rate monitor.

According to a first aspect of the present invention, there is provided a respiration rate monitor for measuring the respiration rate of a body, the monitor comprising at least one accelerometer attachable to an upper part of a body and arranged to provide an accelerometer output signal which is related to the respiration rate of that body, and means for determining the said respiration rate from the or each accelerometer output signal and for providing a monitor output indicative of respiration rate, comprising a baseline filter for removing artefacts of the accelerometer output signal which are not a result of respiration of the body, so as to provide a removed baseline signal.

The respiration rate monitor of the invention, when mounted upon the upper part of a body, such as a chest wall of a patient, detects movements of the upper body part synchronous with inhalation and exhalation. From these, the respiration rate of the patient can be determined. The monitor is non-invasive and may be relatively small. Thus, the wearer will, typically, be less aware of its presence than with prior art respiration rate monitors and hence will tend to affect the patient's breathing patterns less.

Preferably, the monitor comprises means for determining the said respiration rate from the or each accelerometer output signal. These may be local to the accelerometer or remote therefrom, and may be provided in hardware, software or a combination thereof.

It is preferable that the respiration rate monitor includes a low pass filter which is arranged to remove frequencies above a threshold value from the accelerometer output signal. Typically, this threshold value will be determined on the basis of the maximum possible observed respiration rate of the human body.
Optimally, the threshold frequency will be in the range 0.1 to 3 Hz. Although either an analogue or a digital filter could be used, it is preferable to use an analogue filter as this simultaneously limits noise and acts as an anti-aliasing filter.

Accelerometers measure static as well as dynamic acceleration. This means that the output of the sensor is related both to movement and the changes in effective gravity (caused by the angle of the sensor changing with respect to the vertical). This means that the "raw" accelerometer output signal consists of a signal related to chest wall movement caused by respiration, superimposed upon a varying baseline signal. Removing the baseline signal is desirable because it is not really related to respiration and is, therefore, a form of noise, and also because the signal which remains when the baseline is removed can be more accurately digitised because the signal has a smaller amplitude range.

Preferably, the baseline filter comprises means for generating a demand signal representative of an estimated baseline signal, and means for subtracting the said demand signal representative of an estimated baseline signal, and means for subtracting the said demand signal from the accelerometer output signal, or a derivative thereof, so as to produce the removed baseline signal. Although this procedure can be done using analogue or digital techniques, it is most preferable to employ a combined analogue/digital technique. In a particularly preferred embodiment of the invention, therefore, the means for generating the demand signal is controlled on the basis of a digital representation of the removed baseline signal, but generates the demand signal as an analogue signal. The means for subtracting the demand signal from the accelerometer output signal, or a derivative thereof, is then preferably operable to subtract the analogue
demand signal from an analogue accelerometer output signal or a derivative thereof.

The monitor of the invention further preferably comprises a bandpass filter arranged to extract, from the removed baseline signal, those frequencies associated with the respiration rate of the body, and to output a bandpass filtered signal containing those extracted frequencies. The currently preferred solution is to use a digital bandpass filter.

The monitor further preferably comprises means for determining the power spectrum of the bandpass filtered signal. One solution is to take a Fourier transform of the digitised bandpass filtered signal. Most preferably, however, a power spectrum is constructed using autoregressive (AR) modelling. Frequency spectra constructed using AR coefficients provide more accurate information about the predominant frequencies in the signal.

Where the frequency spectrum has been obtained using a Fourier transform, it is preferable that a search and interpolate algorithm is employed to indicate a dominant frequency. This dominant frequency is the frequency of respiration. By contrast, if the particularly preferred technique of determining AR coefficients is employed, then the spectrum produced by the AR coefficients is preferably searched to locate the dominant frequency therein. Most preferably, the search range is restricted to frequencies corresponding to physiologically plausible rates of respiration.

The determined respiration rate may be output as a digital value. This can be achieved by including data communication means as part of the monitor such that the monitor output is in computer-readable form. Additionally or alternatively, the monitor itself may further comprise display means which is arranged to receive the monitor output, and to provide to a user
an indication of the determined respiration rate. Furthermore, the monitor may comprise an alarm which is preferably arranged to be actuated by the monitor in response to the determined respiration rate falling outside a predetermined range. This predetermined range may be varied depending upon the needs of the patient.

According to a second aspect of the present invention, there is provided a combined heart and respiration rate monitor comprising the respiration rate monitor defined in the first aspect of the present invention but where the accelerometer output signal is further related to the heart rate of the body.

Preferably, the combined heart and respiration rate monitor further comprising means for determining the said heart rate from the or each accelerometer output signal, the monitor output being further indicative of heart rate.

The ability to monitor both respiration and heart rate simultaneously and using the same sensor (accelerometer) is particularly useful. In preferred embodiments, the combined respiration and heart rate monitor processes the (single) accelerometer output signal by passing it through either a single bandpass filter having two pass bands at different frequencies, or, alternatively, by passing the same signal through two separate bandpass filters which have different pass bands. In this manner, a first bandpass filtered signal can be obtained which contains only those frequencies derived from respiration, and a second bandpass filtered signal can be obtained with only those frequencies related to heart rate in it. The two, separate, bandpass filtered signals can be processed further separately to identify respiration and heart rates respectively. These may be displayed separately and/or simply output as different digital
values for storage for transmission to other places.

In a third aspect of the invention, there is provided a heart rate monitor for measuring the heart rate of a body, the monitor comprising: (a) at least one accelerometer attachable to a part of a body and arranged to provide an accelerometer output signal which is related to the heart rate of that body; (b) a low pass filter arranged to filter out those frequencies higher than a threshold frequency which is related to a chosen maximum heart rate; (c) a baseline filter for removing artefacts of the accelerometer output signal which are not a result of heart beats and for generating a removed baseline signal; (d) a bandpass filter arranged to extract from the removed baseline signal, those frequencies associated with the heart rate of the body and to output a bandpass filtered signal containing those extracted frequencies; (e) means for determining a power spectrum of the bandpass filtered signal; and (f) means for searching the determined power spectrum so as to detect the dominant frequency thereof, the dominant frequency being indicative of heart rate; the heart rate monitor providing a monitor output indicative of the thus determined heart rate.

In yet a further aspect of the present invention, there is provided a method of measuring the respiration rate of a body comprising the steps of attaching at least one accelerometer to an upper part of a body, providing an accelerometer output signal which is related to the respiration rate of that body, determining the said respiration rate from the or each accelerometer output signal, providing a monitor output indicative of respiration rate, and removing artefacts of the accelerometer output signal which are not a result of respiration of the body, so as to provide a removed respiration baseline signal.

Further advantageous features of the invention
are set out in the dependent claims appended hereto.

The invention may be put into practice in a
number of ways, and one preferred embodiment will now
be described by way of example only and with reference
to the accompanying drawings in which:

Figure 1 shows, schematically, the monitor of the
present invention located adjacent a patient’s chest
wall and including one or more accelerometers;

Figure 2 shows, again schematically, the
functional steps in the processing of an output signal
from the or each accelerometer of Figure 1; and

Figure 3 shows, in further detail, a part of the
signal processing arrangement of Figure 2.

Figure 1 shows, schematically, a breathing and/or
heart rate monitor embodying the present invention.
The monitor, in the embodiment shown in Figure 1,
comprises an accelerometer arrangement 10 mounted upon
a pad 20. The pad 20 is typically adhesive on its rear
face so as to allow it to be stuck to a patient’s
skin. The illustrated embodiment is intended to detect
both respiration and heart beats and it has been found
that, for this application, it is most preferable to
locate the accelerometer arrangement upon the sternum
at the bottom of the rib cage 30.

The accelerometer arrangement 10 comprises an
accelerometer such as the Analogue Devices ADXL202 or
ADXL105 accelerometers. The accelerometer is supplied
as an integrated silicon chip provided with an
analogue output and a duty cycle. Although either of
these signals can be used, the analogue signal is
currently preferred as subsequent signal processing is
partly carried out in the analogue regime. If the duty
cycle were to be used, however, this would be
converted into a digital signal immediately and all
subsequent processing would be carried out digitally.

Still referring to Figure 1, the analogue output
signal from the accelerator arrangement 10 is passed
to a monitor/processor 40 which typically comprises a screen 50, a breathing rate indicator 60 and a heart rate indicator 70. An alarm 80, which may be audible, visual or both, may also be provided. A digital data output 90 from the monitor/processor 40 may also be provided.

Referring now to Figure 2, the processing of the accelerometer output signal will now be described. Figure 2 is also a schematic diagram illustrating the functions that are carried out in the processing of the accelerometer output signal. As shown in Figure 1, most of the signal processing is carried out remote from the patient’s body using software; this minimises the amount of hardware that must be attached to the patient’s body. All of the processing could, of course, be carried out within the accelerometer arrangement 10 which is mounted upon the patient’s body, and in that case most likely in hardware. Nevertheless, the functional steps in the processing of the accelerometer signal would in preference be similar in each case.

Following generation of the analogue accelerometer output signal from the accelerometer arrangement 10, this signal is sent firstly to a low pass filter 100. The low pass filter 100 acts as a noise removal device. Those frequencies in the signal above a threshold frequency are removed by the low pass filter 100. The threshold frequency is determined in dependence upon the physiological parameter being measured. If the monitor is intended to measure only breathing rate, then frequencies above a few Hz can safely be discarded since respiration rates above this are physiologically implausible. If heart rate is to be monitored, or both are to be monitored, then a slightly higher cut-off frequency will be necessary, perhaps in the range 5-10 Hz.

Depending upon whether the accelerometer output
is analogue or digital, the low pass filter can likewise be either analogue or digital. For digital signals, either a finite impulse response (FIR) or an infinite impulse response (IIR) filter may be used.

In the preferred embodiment where the accelerometer output signal is analogue, the low pass filter is likewise analogue. This provides the advantage that high frequency noise is limited and, at the same time, the low pass filter 100 acts as an anti-aliasing filter.

The next stage in the signal processing is to remove the “baseline” in the low pass filter accelerometer output signal. Accelerometers measure static as well as dynamic acceleration. This means that the output of the sensor is related both to movement and to changes in effective gravity (caused by the angle of the sensor changing with respect to the vertical). This means that the movement signal, which it is desirable to analyse, is superimposed on a varying baseline signal as the patient moves. There are two reasons for wishing to remove this baseline. Firstly, it is “noise” in the sense that it is not really related to either respiration or heart rate. Secondly, by removing the baseline, the remaining, information-bearing signal can be digitised more accurately because less of the digitisation range has to be used to allow for changes in the baseline.

Removing the baseline is a challenging problem because respiration itself is a low frequency signal. A number of approaches have been considered. Firstly, a simple analogue high pass filter may be employed to remove the low frequency signal resulting from patient movement. The cut-off frequency is likely to be in the range 0.01 to 0.1 Hz. In this case, the low pass filter 100 and baseline removal 110 could be combined into a single band pass filter that extracts only those frequencies between a fraction of a Hz and
10 Hz.

Alternatively, the analogue output from the low pass filter could be digitised and those frequencies below a fraction of a Hz could then be discarded. In this case, it would be preferable to use a digital band pass filter on a digital output from the accelerometer arrangement 10 instead.

In the analogue domain, a better approach is to estimate the signal baseline and subtract it from the "raw" signal which is an output of the low pass filter 100. The estimate on the signal baseline may be generated using an analogue integrator. In the digital domain, a digital filter may be used to generate a digital estimate of the signal baseline which again may be subtracted from the raw output from a (digital) low pass filter.

The currently preferred approach, however, is to use a combination of analogue and digital techniques. The control loop for the preferred baseline removal arrangement 30 is shown in more detail in Figure 3. The baseline is removed by subtracting an analogue estimate of the baseline from the raw analogue signal arriving from the low pass filter 100. The subtraction of one analogue signal from another is a procedure which will be familiar to those skilled in the art and may, for example, be carried out using a suitably configured operational amplifier. This procedure is shown in block 112 in Figure 3.

The resultant analogue signal in which the baseline has been removed is passed to an analogue-to-digital converter 114 which digitises the analogue signal. As explained previously, once the baseline has been removed the dynamic range of the resultant respiration and/or heart signal is significantly reduced which increases the available accuracy of digitisation. The digitised respiration/heart rate signal is then passed to the next stage in the signal
processing. It is also, however, passed to a baseline monitor 115 which, in combination with a controller 116, tracks the baseline and generates a digital control or demand signal. This control signal is used in turn to generate the estimated baseline using signal generator 118. It is this estimated baseline which is subtracted from the input signal to the baseline removal algorithm 110. In other words, the subtraction is carried out in the analogue domain, but the estimate of the baseline which is to be subtracted is controlled from the digital domain.

Referring once more to Figure 2, the output from the baseline removal algorithm 110 is sent to separate band pass filters 120 and 160. The band pass filter 120 is configured to extract frequency from the digital signal that is now put at the baseline removal algorithm 110 that are associated with respiration. It is preferable, since the output of the baseline removal algorithm 110 is digital, that band pass filtering by the band pass filter 120 is also carried out in the digital domain.

After the frequencies associated with respiration have been extracted, the resultant signal is further processed to determine its power spectrum. This is carried out at step 130 of Figure 2. Determination of a power spectrum is traditionally carried out by taking a Fourier transform of the digitised signal. However, a discrete frequency spectrum estimation process such as a fast Fourier transform (FFT) is inefficient in the present case, where the expected frequency spectrum contains only one large peak corresponding with the patient breathing rate. In order to estimate the frequency of that peak accurately, a large number of frequency bins would be required, of which the majority would remain unused. A further problem with FFT’s is the difficulty in distinguishing close spectral peaks which is relevant
in the present case where heart rate and breathing rate are typically of the same order of magnitude.

Therefore, in presently preferred embodiments, an autoregressive (AR) model is employed, since this model's transfer function should contain poles. Autoregression allows that part of the spectrum where the breathing signal may exist to be zoomed in upon, such that a continuous frequency spectrum estimate may be generated in this range. For the calculation of AR coefficients, the Burg algorithm (or maximum entropy method) would appear to provide more accurate results. The number of coefficients is adjusted empirically to provide an acceptably accurate spectrum estimate without requiring excessive processing time.

Once the power spectrum has been determined, the respiration rate is determined at step 140 in Figure 2. If the frequency spectrum has been obtained at step 130 using a Fourier transform, then a traditional search and interpolate algorithm can be used to indicate the dominant frequency, which is the frequency of respiration. If, as is preferred, AR coefficients have been used, then the spectrum produced by the AR coefficients is searched to find the dominant frequency. By simply searching for peaks in the appropriate frequency range (somewhere around 1 Hz for human breathing rates) the problem of DC offset (which appears as a large peak at zero Hz) can be avoided.

The determined respiration rate can then be utilised in a number of ways. The breathing rate may, as shown in Figure 1, be displayed as a number on the monitor/processor 40, and/or the breathing pattern may be displayed upon the screen 50. If the breathing rate as determined by the algorithm of Figure 2 is outside clinically acceptable boundaries (such as, for example when apnoea or cessation of breathing occurs) then the alarm 80 may be activated. The signal may also be
output in digital form using a communications protocol, for example on a digital bus, using a serial protocol such as RS-232 or USB, across a LAN or via wireless technology. In addition to constantly updated breathing rate values, trends may be displayed as well or instead. Furthermore, the data output 90 may be stored to a storage medium such as a computer hard or floppy disk.

In addition to determination of breathing or respiration rate, heart rate may also be monitored. Referring again to Figure 2, the digital output from the baseline removal algorithm 110 is split into two and the second of the two signals is passed through a second band pass filter 160 which is tuned to pass only those signals within a range of physiologically plausible heart rates. Typically, these rates are somewhat higher than the range of physically plausible breathing rates. After filtering with the second band pass filter 160, the power spectrum is once again determined using AR coefficients at step 170 and the heart rate is ultimately determined, again by identifying the dominant peak in the determined power spectrum, at step 180. The determined heart rate may be displayed on the monitor/processor 40 separately, as a number, in the heart rate indicator 70. Heart rate data can likewise be output from the monitor/processor 40.

Whilst one specific embodiment has been described, it is to be understood that many variations to the foregoing are contemplated. Firstly, although a single accelerometer arrangement 10 is shown affixed to a location adjacent a chest wall at the base of the sternum, there are a number of other places that it could be located. Moreover, it is not essential that the signal which is output from the accelerometer arrangement 10 is used to determine both heart rate and breathing rate. It may be preferable, if it is
desirable to monitor both heart and breathing rates, to use separate monitors, perhaps located at different locations on the body.

Where the signal processing is carried out remote from the accelerometer arrangement 10, the physical dimensions of the arrangement itself may be relatively small. For example, pads 20 which are not much larger than ECG pads could then be employed. However, the monitor as shown in Figure 1 is more suited to hospital environments. All of the signal processing could, by contrast, be carried out using dedicated software or hardware built into the accelerometer arrangement 10 such that the output of the accelerometer arrangement was no longer a raw signal but is instead a processed signal which directly indicates the breathing and/or heart rate. This processed information may be sent either via a wire or wirelessly to a small wrist monitor. The wrist monitor may display the actual heart and/or breathing rate, or may simply have an alarm which is triggered when the heart and/or breathing rate moves outside an acceptable range.

Finally, it is to be understood that, whilst a single accelerometer is currently preferred, two or more accelerometers could equally be used. The advantage of multiple accelerometers is that the outputs from each can be combined to improve noise rejection caused by movement of the patient.
CLAIMS:

1. A respiration rate monitor for measuring the respiration rate of a body, the monitor comprising:
   at least one accelerometer attachable to an upper part of a body and arranged to provide an accelerometer output signal which is related to the respiration rate of that body; and
   means for determining the said respiration rate from the or each accelerometer output signal and for providing a monitor output indicative of respiration rate, comprising:
   a baseline filter for removing artefacts of the accelerometer output signal which are not a result of respiration of the body, so as to provide a removed baseline signal.

2. The monitor of claim 1, in which the means for determining the said respiration rate further comprises a low-pass filter arranged to remove frequencies above a threshold value from the said accelerometer output signal.

3. The monitor of claim 2, in which the low-pass filter is an analogue filter.

4. The monitor of any of claims 1-3, in which the baseline filter comprises means for generating a demand signal representative of an estimated baseline signal, and means for subtracting the said demand signal from the accelerometer output signal, or a derivative thereof, so as to produce the said removed baseline signal.

5. The monitor of claim 4, in which the means for generating the demand signal is controlled on the basis of a digital representation of the removed
baseline signal, but generates the said demand signal as an analogue signal, the means for subtracting the said demand signal from the accelerometer output signal or a derivative thereof being operable to subtract the said analogue demand signal from an analogue accelerometer output signal or derivative thereof.

6. The monitor of either of claims 4 or 5, further comprising a bandpass filter arranged to extract, from the removed baseline signal, those frequencies associated with the respiration rate of the body and to output a bandpass filtered signal containing those extracted frequencies.

7. The monitor of claim 6, in which the baseline filter is arranged to generate the removed baseline signal as a digital signal, and in which the bandpass filter is a digital signal filter.

8. The monitor of claim 6 or claim 7, further comprising means for determining a power spectrum of the bandpass filtered signal.

9. The monitor of claim 8, in which the means for determining the respiration rate includes means for searching the power spectrum of the bandpass filtered signal to detect the dominant frequency thereof.

10. The monitor of claim 8 or claim 9, in which the means for determining the power spectrum includes a Fourier transform algorithm.

11. The monitor of claim 8 or claim 9, in which the means for determining the power spectrum includes an autoregressive algorithm.
12. The monitor of any preceding claim, further comprising a communications device arranged to provide the said monitor output in computer-readable form.

13. The monitor of any of claims 1 to 12, further comprising display means arranged to receive the monitor output and to provide to a user an indication of the determined respiration rate.

14. The monitor of claim 12 or claim 13, further comprising an alarm, the monitor being arranged to actuate the said alarm in response to the determined respiration rate falling outside a predetermined range.

15. A combined heart and respiration rate monitor comprising the respiration rate monitor of any preceding claim, wherein the accelerometer output signal is further related to the heart rate of the said body.

16. The combined heart and respiration rate monitor of claim 15, further comprising means for determining the said heart rate from the or each accelerometer output signal, the monitor output being further indicative of heart rate.

17. The combined heart and respiration rate monitor of claim 16, when dependent upon claim 6 or claim 7, in which the bandpass filter is further configured to extract from the accelerometer output signal or a derivative thereof those frequencies associated with the heart rate of the body and to output a second bandpass filtered signal containing those extracted frequencies relating to the said heart rate.
18. The combined heart and respiration rate monitor of claim 16 when dependent upon claim 13, in which the display means is arranged to display an indication of both the respiration and heart rates of the body.

19. A heart rate monitor for measuring the heart rate of a body, the monitor comprising:
   (a) at least one accelerometer attachable to a part of a body and arranged to provide an accelerometer output signal which is related to the heart rate of that body;
   (b) a low pass filter arranged to filter out those frequencies higher than a threshold frequency which is related to a chosen maximum heart rate;
   (c) a baseline filter for removing artefacts of the accelerometer output signal which are not a result of heart beats and for generating a removed baseline signal;
   (d) a bandpass filter arranged to extract from the removed baseline signal, those frequencies associated with the heart rate of the body and to output a bandpass filtered signal containing those extracted frequencies;
   (e) means for determining a power spectrum of the bandpass filtered signal; and
   (f) means for searching the determined power spectrum so as to detect the dominant frequency thereof, the dominant frequency being indicative of heart rate; the heart rate monitor providing a monitor output indicative of the thus determined heart rate.

20. The heart rate monitor of claim 19, in which
the low-pass filter is an analogue filter.

21. The heart rate monitor of claim 19 or claim 20, in which the baseline filter comprises means for generating a demand signal representative of an estimated baseline signal, and means for subtracting the said demand signal from the accelerometer output signal, or a derivative thereof, so as to produce the said removed baseline signal.

22. The heart rate monitor of claim 21, in which the means for generating the demand signal is controlled on the basis of a digital representation of the removed baseline signal, but generates the said demand signal as an analogue signal, the means for subtracting the said demand signal from the accelerometer output signal or a derivative thereof being operable to subtract the said analogue demand signal from analogue accelerometer output signal or derivative thereof.

23. The heart rate monitor of any of claims 19 to 22, in which the baseline filter is arranged to generate the removed baseline signal as a digital signal, and in which the bandpass filter is a digital signal filter.

24. The heart rate monitor of any of claims 19 to 23, in which the means for determining the power spectrum includes a Fourier transform algorithm.

25. The heart rate monitor of any of claims 19 to 23, in which the means for determining the power spectrum includes an autoregressive algorithm.

26. The heart rate monitor of any of claims 19 to 25, further comprising a communications device.
arranged to provide the said monitor output in computer-readable form.

27. The heart rate monitor of any of claims 19 to 26, further comprising display means arranged to receive the monitor output and to provide to a user an indication of the determined heart rate.

28. The heart rate monitor of claim 26 or claim 27, further comprising an alarm, the monitor being arranged to actuate the said alarm in response to the determined heart rate falling outside a predetermined range.

29. A method of measuring the respiration rate of a body comprising the steps of:
attaching at least one accelerometer to an upper part of a body;
providing an accelerometer output signal which is related to the respiration rate of that body;
determining the said respiration rate from the or each accelerometer output signal;
providing a monitor output indicative of respiration rate; and
removing artefacts of the accelerometer output signal which are not a result of respiration of the body, so as to provide a removed baseline signal.

30. The method of claim 29, further comprising:
removing frequencies above a threshold value from the said accelerometer output signal.

31. The method of claim 29, in which the step of removing artefacts comprises generating a demand signal representative of an estimated baseline signal, and subtracting the said demand signal from the accelerometer output signal, or a derivative thereof,
so as to produce the said removed baseline signal.

32. The method of claim 31, in which the step of generating a demand signal comprises receiving a digital representation of the removed baseline signal and generating an analogue demand signal on the basis thereof, and in which the step of subtracting the signals comprises subtracting the analogue demand signal from an analogue accelerometer output signal or derivative thereof.

33. The method of any of claims 29 to 32, further comprising extracting, from the removed baseline signal, those frequencies associated with the respiration rate of the body and outputting a bandpass filtered signal containing those extracted frequencies.

34. The method of claim 33, further comprising determining a power spectrum of the bandpass filtered signal.

35. The method of claim 34, further comprising searching the power spectrum of the bandpass filtered signal to detect the dominant frequency thereof.
FIG. 2.

ACCELEROMETER ARRANGEMENT

LOW PASS FILTER

BASELINE REMOVAL ALGORITHM

BAND PASS FILTER

DETERMINE POWER SPECTRUM

DETERMINE RESPIRATION RATE

BAND PASS FILTER

DETERMINE POWER SPECTRUM

DETERMINE HEART RATE

OUTPUT, DISPLAY, STORE, PRINT, etc.