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A device for 24 hour ambulatory monitoring of abdominal girth using inductive plethysmography

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Abstract

Inductive plethysmography (IP) sensors and oscillator modules were assessed for their potential use in the ambulatory monitoring of abdominal girth in subjects with irritable bowel syndrome (IBS) in order to objectively quantify their bloating symptoms. A dedicated microprocessor data logger was designed to record over 24 h the frequency output of IP oscillators connected to a belt around the subject's lower abdomen. Posture was also recorded via tilt switches (standing, sitting and lying).

The system was separately calibrated by placing the belts around a variable rectangular phantom and measuring the frequency of oscillation. A theoretical geometric model was devised to convert measured frequency into circumference and account for changes caused by variations in shape. Using the calibration factors, it was found that the circumference of a circular phantom could be measured accurately (mean difference 1.27 cm and SD 0.25 cm).

The system has been tested over 24 h with 20 volunteers. Movement introduced variations in measured girth larger than those found during periods of non-movement during sleep.

We conclude that IP promises to be a useful and quantitative tool suitable for ambulatory monitoring of abdominal girth, a hitherto relatively unexplored symptom of IBS.

Keywords: IBS, irritable bowel syndrome, bloating, plethysmography

1. Introduction

Abdominal bloating is one of the principal symptoms of irritable bowel syndrome (IBS) (Manning *et al* 1978), but it is probably the most poorly understood. It has been rated by

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patients as the most uncomfortable symptom of their condition (Lembo *et al* 1999), considerably more so than even abdominal pain. There is even some difference of opinion amongst clinicians regarding whether the symptom actually exists or is merely reported by patients who feel bloated for other reasons, without any real change in girth. There has been very little quantitative investigation of bloating associated with IBS. Such measurements (Maxton *et al* 1991) have been conducted under unnatural, possibly stressful conditions, usually in the presence of the clinicians whom the patient wishes to convince of the nature of his or her symptoms, with all the attendant uncertainties which can arise from that situation.

A quantitative ambulatory measurement system could provide independent data for the first time that shows patterns of bloating as the patients went about their normal lives away from a clinic. This could allow investigation into the causes of bloating (still largely the subject of speculation) and the effects of pharmaceuticals designed to alleviate the symptom.

We wished to devise an objective ambulatory system to measure unobtrusively the circumference of a patient's abdomen over a full 24 h period. Inductive plethysmography (IP) has been extensively used in non-ambulatory settings for quantitative monitoring of respiration, with varying degrees of success. Researchers have variously found it to be a quantitative (Stefano *et al* 1986), non-quantitative (Ballard *et al* 1988) or most often a 'semi-quantitative' (Caretti *et al* 1994) method for monitoring respiration. In almost all cases, calibration is performed *in vivo* against a standard measuring device, e.g., a spirometer. Lack of accuracy under these circumstances can be attributed either to the measuring system or to an imperfect relation of breathing to movements in the chest wall, and other physiological factors. Konno and Mead (1967) using conventional techniques to measure surface movements found that ventilation could be measured indirectly by treating the chest wall as two compartments: the rib cage and the abdomen. This was limited, however, to quiet breathing in a single posture. Changes in posture or distortions in the chest wall introduced errors. The subsequent respiration work is built upon this early study and has reproduced some of the inaccuracies found by Konno and Mead. Our intention was not, however, to find an indirect measure of a physiological parameter, but to directly monitor the bulk of the lower abdomen, whatever the cause. IP therefore promised to be a solution, possibly without some of the problems encountered with the respiration work.

There remains the question of the intrinsic accuracy of IP in reflecting the cross-sectional area enclosed by the elasticated belts. Fewer researchers have investigated this. The patent holders (Watson *et al* 1982) describe it as a system with output proportional to the area enclosed by the belt loop. Martinot-Lagarde *et al* (1988) examined this claim and found that the inductance of the belt did indeed vary with area, but with some other factors influencing the output, most notably shape. Watson *et al* (1988) subsequently looked at the accuracy of IP in the light of this research and found a base-line shift in output between rectangular and curved shapes, but that for both, output varied continuously with area. They concluded that suitably calibrated IP could accurately measure cross-sectional area within the physiological range.

Alternate direct length measurement techniques were considered; however, IP was advantageous owing to the fact that it relied upon a weakly elasticated belt as a sensor. This allows the sensor to conform well to the abdomen during both expansion and contraction. Since IBS is a psychologically related condition, we believed it is important that the subject be directly as unaware as possible of the presence of the sensors. IP sensors are extremely comfortable and unobtrusive. These mechanical properties of the IP sensor were difficult to reproduce with alternate sensors.

2. Methods

2.1. Apparatus

The IP system incorporates a wire loop sewn into a stretchable belt as an inductor in a resonant Colpitts oscillator circuit (Watson *et al* 1982). The oscillator is housed in a small plastic box physically close to the belt, and outputs a variable frequency square wave signal centred around approximately 450 kHz. This signal is normally fed into a mains operated frequency to voltage converter and filter for analysis. Clearly this equipment is unsuitable for an ambulatory study. The oscillator, however, is physically small and powered by a small dc voltage (2–6 V) at a low current (<1 mA). Our intention was to use only this part of the system and a dedicated microprocessor based data logger to record the IP signal.

A small (weight 90 g, size 72 mm $\times$ 55 mm $\times$ 18 mm), portable, battery operated data logger was designed based around the microchip PIC16F84 microcontroller. Non-volatile serial EEPROM (16 kbytes) was included for data storage. The logger was powered from two half-AA style lithium thionyl chloride cells in series. These have a nominal voltage of 3.6 V each and capacity 1000 mAh. The power was regulated to 5.0 V to provide a stable supply to the microcontroller, EEPROM and IP oscillators. Total current consumption was approximately 2 mA, giving a battery life of up to 500 h. The hardware element of the data logger was kept simple and physically small. Built into the software residing on flash EEPROM on the microcontroller chip were routines to communicate with the external serial memory via an I^2C bus, and with a PC via an RS232 link for downloading data and for calibration.

The logger's primary purpose was to record over 24 h, the average frequency of the oscillator once per minute. Since the oscillator output was found to be logic level compatible, no analogue conversion was necessary and the pulse counting capabilities of the PIC16F84 were utilized to monitor the number of cycles of the oscillator in fixed times. The accuracy of this frequency measurement was determined only by the accuracy and stability of the 4 MHz crystal clock which drove the microcontroller.

Two separate oscillators were housed in the oscillator box, normally used for the chest and abdomen signals in respiration monitoring. Whilst we required only one belt, there was a small drift in the oscillator frequency output with temperature, and so the free running output of the second oscillator, with no belt connected, was also recorded. This drifted only with temperature and so with suitable calibration provides a means of correcting the primary girth signal for temperature drift. Each of the two frequencies was measured over successive 30 s intervals.

Previous researchers have found that the signal recorded using IP to monitor respiration shows significant base-line shift with changes in posture (Zimmerman *et al* 1983). It was expected that similar shifts would be encountered in this case and so two sealed mercury tilt switches which open when within 45° of the vertical were also attached to the data logger. These were fixed to the side of the subject's chest and thigh, allowing posture (standing, sitting and lying) to be monitored and recorded simultaneously with the IP girth signal. The switches were also connected directly to the microcontroller's digital inputs. The microcontroller was programmed to monitor these inputs, and filter out changes of less than 30 s duration, to eliminate jitter caused by movement and vibration.

2.2. Theory

The frequency of a Colpitts oscillator is inversely proportional to the square root of the inductance included in the circuit. We required therefore an expression relating the length of a wire turned into a loop and its inductance.

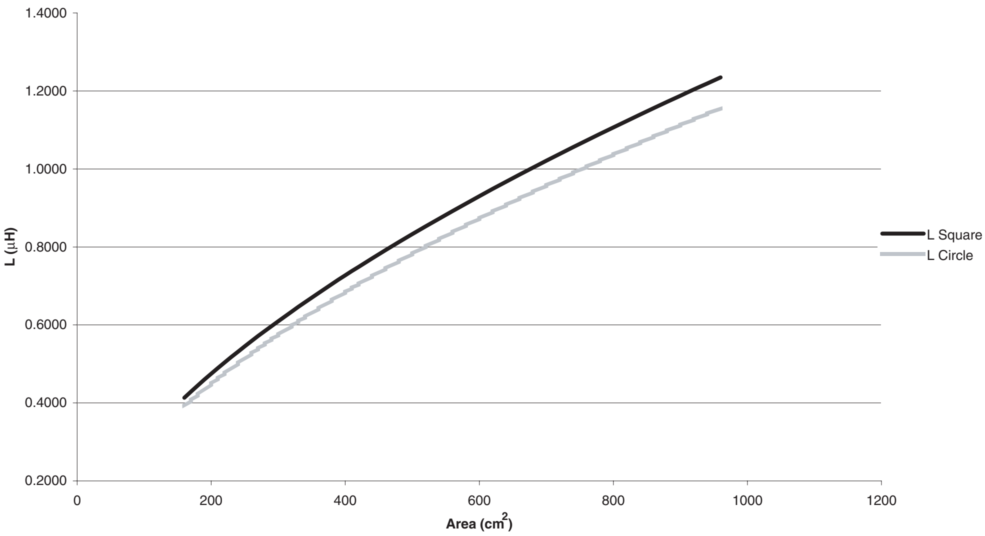

Figure 1. Inductance versus area for curved and non-curved shapes.

Bashenoff (1927) produced a general equation (1)

$$L = 2 \times 10^{-7} l \left[\ln\left(\frac{2l}{\rho}\right) - \left(2\ln\left(\frac{l}{\sqrt{S}}\right) + \varnothing\right) + \frac{\mu_r}{4} \right] \tag{1}$$

giving inductance L (H) in terms of l, the length of the loop (m), S, the area enclosed by the loop (m^2), ρ, the radius of cross section of the wire (m), and $\varnothing$, a small dimensionless shape-dependent correction factor. μ_r is the relative permeability of the wire. $\varnothing$ is known analytically for various shapes, and it has been shown (see Grover 1962) that for shapes where $\varnothing$ is not known, it may be interpolated from tables of similar shapes, using the ratio $l/\sqrt{S}$. For a given shape, $l/\sqrt{S}$ is fixed and inductance therefore depends only on the length of the inductor. Grover gives the example of a 100 cm long wire bent into an ellipse of eccentricity $1/\sqrt{2}$. From this, $l/\sqrt{S}$ is calculated and $\varnothing$ interpolated from a table of $\varnothing$ versus $l/\sqrt{S}$ for regular polygons of varying numbers of sides. The value calculated via this method (6.81 μH) is close to the analytic value for a circle of the same length: 6.91 μH. The importance of this result lies in the fact that results obtained using the approximate model are close to those using the same length of wire bent only into a broadly similar shape and suggests that this model may be used to predict the length of an inductor from the measured inductance. Equation (1) is strictly valid in the case of a planar inductor and l represents the physical length of the wire used. In the case of IP belts, the inductor deviates from a single plane because of the zigzag pattern introduced to allow expansion. Also, although the length of the belt changes, the wire itself does not stretch and remains a fixed length. For the purposes of this model, it is assumed that the effects of the zigzag pattern may be neglected, and the inductance of the belt varies as though the length of the inductor is varying with the length of the belt.

Applying equation (1) closely reproduces the empirical results found by Watson *et al* (1988). The inductance was calculated for a range of cross-sectional areas of squares and circles (160–960 cm^2) and the results are plotted in figure 1. As can be seen, inductance rises with cross-sectional area, with a weak negative curvature, and is larger for rectangular shapes than the curved ones. These are all features found empirically by Watson *et al* (1988).

In order to apply equation (1) in a physiological setting, a model of the shape of the lower abdomen was required. Martinot-Lagarde *et al* 1988 modelled the upper and lower abdomen as an ellipse and we have used this as our geometric model. Before attaching the belt to a subject, the anterior–posterior (AP) and lateral diameters of their abdomen were measured using anthropometric callipers. These two values defined an ellipse from which an initial perimeter (l) and area (S) could be calculated, giving $l/\sqrt{S}$ and allowing $\varnothing$ to be interpolated from tables.

2.3. Calibration

2.3.1. Temperature and stability. First, the belt was placed with the oscillator box and data logger around a series of differently sized phantoms inside a temperature controlled enclosure together with a platinum resistance probe. The temperature was raised to approximately 40 °C and allowed to cool to room temperature. The frequencies of the two oscillators were continuously monitored and recorded on a PC, as was the temperature via the resistance probe. The two measured frequencies were fitted using a least-squares method to give a measure of the drift in the girth signal against the drift in the temperature signal. If the frequencies of the two oscillators were simultaneously recorded, all measurements of the girth frequency could be corrected to a standard temperature, arbitrarily chosen to be 30 °C. This correction was performed on all data acquired during subsequent calibration and ambulatory acquisition.

2.3.2. Girth. A motorized computer graph plotter was adapted for the purpose of calibrating the IP belts. Wooden peg boards with spaced holes to accept dowel rods were fixed to one end of the plotter and to the moveable arm. Inserting two dowels in each of these boards produced a rectangle around which the belt was placed. The rectangle could easily be increased or decreased in size along one axis in small movements by moving the arm under computer control. The minimum step size of the arm movement was 0.1 mm. Simultaneously, the data logger measured the frequency output of the oscillators and transmitted these values to the computer via the RS232 link. This facilitated a semi-automatic computer controlled calibration method. Four belts ranging in size from 28 in. to 40 in. were separately calibrated using this method. In each case, the belt was placed around the frame and the initial dimensions of the rectangle were measured with a tape. Under computer control, the frequency of oscillation was measured, and then the frame changed in size by a known amount and the process was repeated. As the dimensions of the frame were known in each case, the theoretical inductance (L_T) of the belt can be calculated using equation (1). The frequency of oscillations for any Colpitts oscillator is proportional to $1/\sqrt{L}$, and so there should be a single linear relationship between frequency and $1/\sqrt{L_T}$ for different sizes and shapes. Plotting frequency against $1/\sqrt{L_T}$ provides both a check of the model, and calibration data for converting the frequency measured during the ambulatory studies to inductance. Using this inductance and equation (1), the length of the belt is calculated.

2.4. Pilot study

Twenty healthy female volunteers were recruited to take part in a pilot study to validate the use of IP as a quantitative ambulatory measure of abdominal girth. Their initial girth was measured with a tape measure at the level of the umbilicus, and their AP and lateral diameters were measured with callipers for the purpose of validating the elliptical model used. An appropriately sized belt was selected and taped securely to the subject's abdomen at the same level. Care was taken to tape the belt in such a way as to minimize slippage, whilst allowing

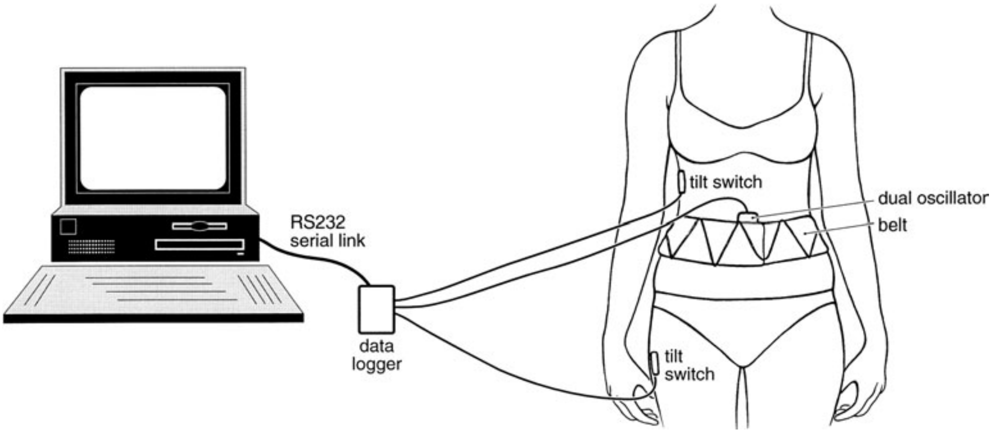

Figure 2. Typical sensor positioning.

the subject as full and free a range of movements as possible. The wire in the belt was attached to the oscillator box, which was taped underneath the belt. A cable ran from this to the data logger, which was worn on a ribbon around the neck. The two tilt switches were taped laterally to the subject's chest and thigh in the vertical position. The typical positioning of the sensors is shown in figure 2.

When powered up, the data logger entered a test mode whereupon two LEDs on the front panel indicated the state of the two tilt switches. This allowed their positioning to be checked by asking the subject to stand, sit and lie down. The positioning was adjusted if necessary and data acquisition was initiated by pushing a small button recessed into the front of the logger.

The subjects then left the laboratory and were asked, as far as possible, to disregard the equipment attached to them and pursue their normal daily activities. They returned to the lab later that day to eat a standard meal at a known time, and again the next morning to have the belt removed, after which the stored data was downloaded via the serial link to a PC. At the start and end of the study, and before and after their evening meal, the subjects were asked to indicate on a visual analogue scale (0–100 mm) their perception of the extent of their own abdominal distention.

2.5. Data acquisition and analysis

The average frequency of the two oscillators was measured and stored successively over 30 s, giving one pair of values per minute. Simultaneously, the two tilt switches were monitored and a 30 s filter applied. The filtered states of the tilt switches were also stored once per minute.

Data was analysed using the initial AP and lateral diameters to define an ellipse, which was assumed constant in shape throughout the study. This allowed $l/\sqrt{S}$ to be calculated using the well-known equation for the area of an ellipse and an approximate equation for its perimeter given by Ramanujan (1914) as follows:

$$C = \pi(3(a+b) - \sqrt{(3a+b)(a+3b)}) \tag{2}$$

where C is the perimeter and a and b are the major and minor semi-axes. $\varnothing$ was interpolated from a look-up table of known values of polygons of various numbers of sides (Grover 1962, p 62). From the measured frequency, the inductance of the belt was calculated using calibration factors, leaving the length of the belt as the only unknown in equation (1), which was solved numerically.

3. Results and discussion

3.1. Temperature and stability

Three different oscillator modules were used, and each was separately calibrated for drift in temperature. The girth signal, with a fixed size belt attached, showed a single-valued parabolic characteristic against the reference temperature signal with a minimum at around 25 °C. Varying the size of the belt shifted the curve up and down without altering its characteristics. The curve in each case was fitted via the least-squares method to a quadratic equation and this data was used to correct the measured frequency of the girth signal to the frequency at a standard fixed temperature (30 °C). The calibration was then repeated for each oscillator, this time applying the temperature correction and calculating the size of the belt using calibration factors to convert measured frequency into physical size (see the next section). Over a range from 25 °C to 35 °C, the average drift for the three oscillators was equivalent to 1.9 mm, the largest being 2.6 mm, and this was deemed acceptably small.

The stability of the system was assessed with respect to the introduction of objects close to or inside the belt loop. A belt was placed around a large plastic bowl. Filling the bowl with water caused no detectable change in output from that measured with the bowl empty, other than what appeared to be caused by the slight deformation of the bowl due to the weight of the water. Similarly, the introduction of metal objects, such as bunches of keys and metal cans, around and inside the belt loop caused no measurable change in the oscillator frequency. Physical proximity to highly radiating devices such as PCs and mobile phones did not affect output either. The presence of very strongly magnetic materials close to or inside the belt loop caused a change in output consistent with a change in belt circumference of up to 3 mm. Highly magnetic materials at close proximity would not normally be encountered, however, and therefore this was considered not significant. This suggests that environmental factors are unimportant, and IP can be used in an ambulatory setting, away from its normal laboratory setting.

3.2. Girth

Each oscillator was separately calibrated with each of the sizes of belts used. A typical set of recorded values is shown in figure 3, which depicts the measured frequency of the oscillator, corrected to a standard temperature, against $1/\sqrt{L_T}$, where L_T is the inductance calculated from the known dimensions of the rectangular phantom. The calibration was repeated a number of times with phantoms of different fixed widths, whilst each time the length was varied under computer control. The expected linear relationship between frequency and $1/\sqrt{L_T}$ is depicted in this graph. As a check of the calibration and the mathematical model, the oscillator frequency was measured when the belt was placed around three differently sized circular phantoms. From the graph, inductance L_T can be obtained from the measured frequency. Using this, the length l of the belt is calculated using equation (1) and the assumption of a circular phantom. The circumferences measured using the two techniques are tabulated in table 1.

As a visual check, the frequency is plotted against $1/\sqrt{L_T}$ on the same graph as the calibration data. As can be seen, these points lie close to the line of calibration data points,

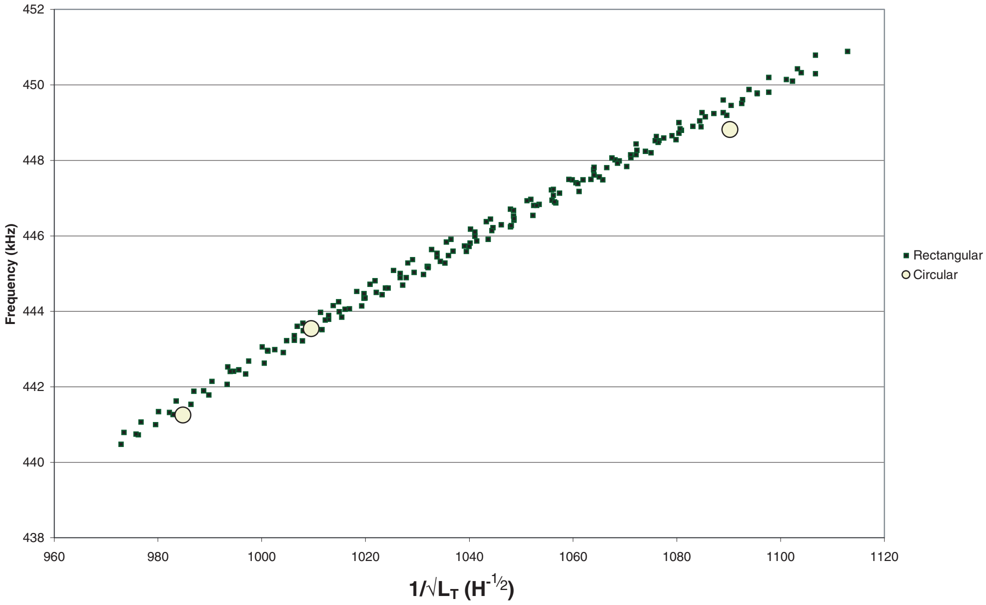

Figure 3. Frequency versus $1/\sqrt{L_T}$: calibration data.

Table 1. Comparison of phantom size measured with a tape and IP.

Phantom	Measured *C* (tape)/cm	Measured *C* (belt)/cm
1	84.2	85.5
2	97.2	95.7
3	99.8	100.8

and the belt-measured circumferences of the phantom show good agreement with the tape-measured values (average absolute difference 1.27 cm and SD 0.25 cm).

In order to assess the utility of this technique, it is necessary to consider what represents a clinically significant change in girth and whether this implementation of IP is capable of measuring that change. Very little quantitative research has been undertaken into bloating associated with IBS. Maxton *et al* (1991) found that between the morning and afternoon IBS, subjects bloated on average 3.5 cm in the lying posture and 4.0 cm whilst standing. Since this is the group of patients who report bloating as a symptom, it is reasonable to say that clinically significant bloating is of this order. Analysis of calibration data revealed that the average change in frequency for a change in belt length of 1 mm was 45.2 ± 3.1 Hz. Frequency was measured by counting the number of signal cycles in a fixed time (30 s). The accuracy of this fixed time was determined only by the stability of the microcontroller's crystal oscillator. In 30 s, a change of over 1300 counts would be expected from a frequency change of 45.2 Hz, suggesting that the resolution of the measurement is a good deal less than 1 mm.

3.3. Pilot study

The main findings from this study are summarized elsewhere (Lewis *et al* 2001). For each of the subjects, their abdominal circumference was estimated from their measured AP and

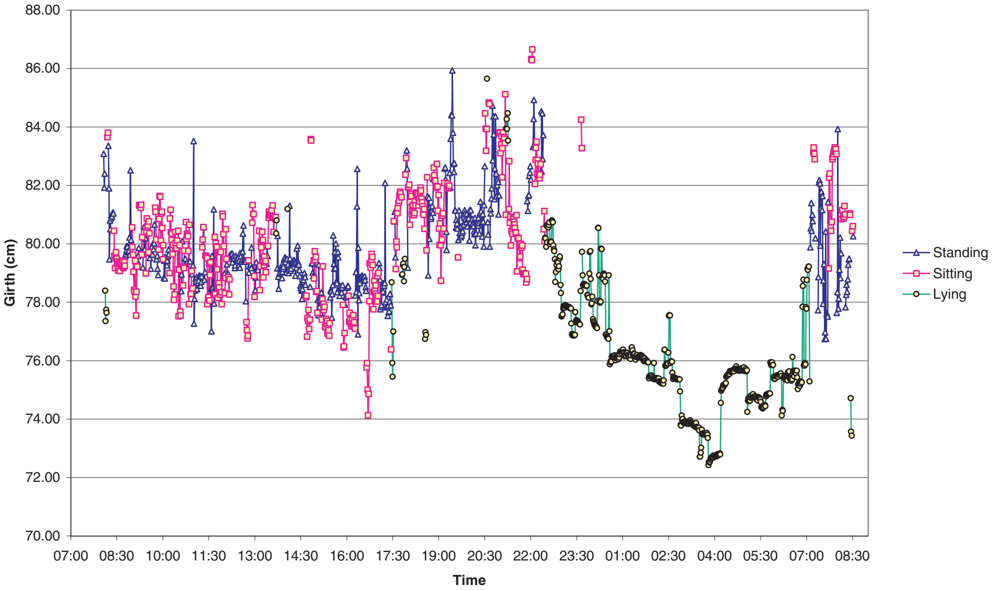

Figure 4. Girth data obtained from a 24 h ambulatory study.
(This figure is in colour only in the electronic version)

lateral diameters using an elliptical model. The ratio of the measured diameters was used to define an ellipse of fixed shape. The geometry of this ellipse allowed the calculation of $l/\sqrt{S}$ and estimation of $\varnothing$ for use in equation (1). These values were used for conversion of the frequency measured during the subsequent 24 h trial to abdominal circumference. The shape-dependent factor in equation (1) is $\{2\log_e(l/\sqrt{S}) + \varnothing\}$. This factor showed little variation between the 20 subjects (average 2.56, range 2.51–2.62), even though they had a wide range of initial abdominal circumferences (range 78.5–128 cm). This suggests that in future studies, a single average value of the shape-dependent factor may be assumed, obviating the need first to measure the subject's abdominal diameters using callipers.

Figure 4 shows a typical trace from one of the 24 h trials. The data are separated into the three postures, and it can clearly be seen that a significant base-line shift occurs when the subject's posture changes from standing or sitting to lying, as may intuitively be expected. Most noticeable from the graph are the occasional sudden changes in measured girth and the relatively large amount of noise present in the data recorded during the day. These changes can be attributed to movement artefacts and slippage of the belt relative to the subject's abdomen. During sleep, minute-by-minute variations are greatly reduced, although occasional sudden changes are still present. For example, during the flat period around 23:00, when the subject appears to be asleep and still, the standard deviation of the girth measurements is 1.3 mm; during an equally long epoch (57 min) around the relatively level period at 10:30 the standard deviation is 8.2 mm. This would indicate that the system is extremely stable and precise, but that movement contributes to artefacts on the recorded signal.

Analysis of the data recorded during this study (Lewis *et al* 2001) showed that there was a significant correlation between the subjects' girth measured prior to the study with a tape and the average of that measured during the first 10 min of the trial with the data logger. By averaging data into epochs before and after meals, a significant increase in measured girth was recorded after the meal. Also, it was found that girth increased on average during the day and returned to normal during the night. Data acquired during the lying posture was significantly lower than during standing and sitting. There was no significant difference between standing and sitting data.

4. Conclusions

It has been shown that IP techniques can be applied continuously in an ambulatory setting over an extended period of time. The length of the belt can be calculated to a good degree of accuracy if minor assumptions are made regarding its shape.

The technique has successfully been applied to the continuous ambulatory measurement of lower abdominal girth. Results were found to be accurate and stable during periods of non-movement. Normal daily movement introduced some random changes. The technique has been evaluated in a pilot study which found that girth measured with IP correlates well with that measured with a tape and that eating increases girth.

This technique promises to be a valuable tool in the ambulatory investigation of IBS and other conditions affecting lower abdominal girth.

References

Ballard R D, Kelly P L and Martin R J 1988 Estimates of ventilation from inductance plethysmography in sleeping asthmatic patients *Chest* **93** 128–33

Bashenoff V J 1927 Abbreviated method for calculating the inductance of irregular plane polygons of round wire *Proc. IRE* **15** 1013–39

Caretti D M, Pullen P V, Premo L A and Kuhlmann W D 1994 Reliability of respiratory inductive plethysmography for measuring tidal volume during exercise *Am. Ind. Hyg. Assoc. J.* **55** 918–23

Grover F W 1962 *Inductance Calculations* (New York: Dover)

Konno K and Mead J 1967 Measurement of the separate volume changes of rib cage and abdomen during breathing *J. Appl. Physiol.* **22** 407–22

Lembo T, Naliboff B, Munakata J, Fullerton S, Saba L, Tung S, Schmulson M and Mayer E A 1999 Symptoms and visceral perception in patients with pain-predominant irritable bowel syndrome *Am. J. Gastroenterol.* **94** 1320–6

Lewis M J, Reilly B, Houghton L A and Whorwell P J 2001 Ambulatory abdominal inductance plethysmography: towards objective assessment of abdominal distension in irritable bowel syndrome *Gut* **48** 216–20

Manning A P, Thompson W G, Heaton K W and Morris A F 1978 Towards positive diagnosis of the irritable bowel *Br. Med. J.* **2** 653–4

Martinot-Lagarde P, Sartene R, Mathieu M and Durand G 1988 What does inductance plethysmography really measure? *J. Appl. Physiol.* **64** 1749–56

Maxton D G, Martin D F, Whorwell P J and Godfrey M 1991 Abdominal distension in female patients with irritable bowel syndrome: exploration of possible mechanisms *Gut* **32** 662–4

Ramanujan S 1914 Modular equations and approximations to π *Q. J. Pure Appl. Math.* **45** 350–72

Stefano J L, Spitzer A R, Baumgart S, Davis J M and Fox W W 1986 Inductive plethysmography—a facilitated postural calibration technique for rapid and accurate tidal volume determination in low birth weight premature newborns *Am. Rev. Respir. Dis.* **134** 1020–4

Watson H, Sackner M A and Stott F D 1982 Method and apparatus for monitoring respiration *US Patent* 4308872

Watson H L, Poole D A and Sackner M A 1988 Accuracy of respiratory inductive plethysmographic cross-sectional areas *J. Appl. Physiol.* **65** 306–8

Zimmerman P V, Connellan S J, Middleton H C, Tabona M V, Goldman M D and Pride N 1983 Postural changes in rib cage and abdominal volume-motion coefficients and their effect on the calibration of a respiratory inductance plethysmograph *Am. Rev. Respir. Dis.* **127** 209–14