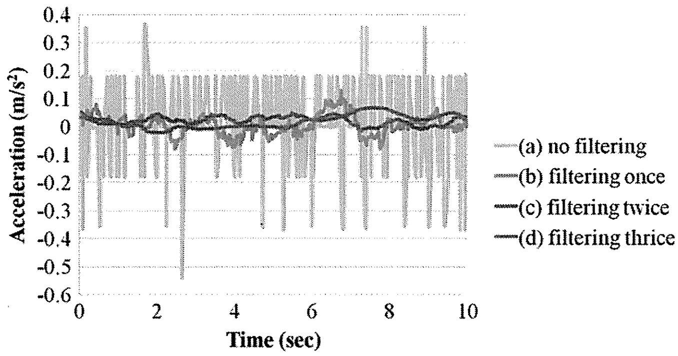

Figure 4. Variations of accelerations which were measured with (a) no filtering, (b) filtering once, (c) filtering twice and (d) filtering thrice.

2.4. Evaluating the accuracy of respiratory monitoring

Respiratory monitoring was performed by the given respiratory motion. In this study, equation (1) was adopted as the reference waveform, where A was kept constant at 5 mm, and τ was varied from 3, to 4 and 5 s. QUASAR™ was used for a physical representation of the thoracic–abdominal wall motion, which has a maximum vertical displacement of 10 mm. This motion was measured five times with the iPod touch® during 3, 4 and 5 s periods of 5 breathing cycles. The respiratory motion data were recorded at a rate of approximately 60 Hz with the iPod touch®.

In order to evaluate the accuracy of respiratory monitoring with the iPod touch®, the absolute means ± SD for the displacement and the period were calculated. All measured waveforms were then shifted to the appropriate point on the reference waveform along the time direction due to the unclear initial time. Because the synchronization between the acceleration sensor of the device and the motion control system of the phantom could not be performed, the initial time was not acquired. Periods were defined as the time from 0 mm amplitude of a cycle to 0 mm of the next cycle. Next, the absolute means ± SD for the displacement and the period were calculated based on the changes in the displacement and the period of the measured waveforms with respect to their corresponding reference waveforms.

3. Results

Figure 4 shows differences in the accelerations measured at (a) no filtering, (b) filtering once, (c) filtering twice and (d) filtering thrice. Figure 4 shows that the greatest changes in acceleration (range, −0.53–0.36 m s^{-2}) occurred during no filtering (a) and gradually smoothed out between (b) and (d). The mean accelerations ± SD and variation coefficients are shown in table 1. The results show clearly that the absolute variation coefficients gradually become smaller between (a) and (d). The mean accelerations ± SD of (c) and (d) were almost the same because the root mean squares (which were calculated from results for the mean accelerations ± SD) for both of them were 0.0029 m s^{-2}. Figure 5 shows the effects of the low-pass filter for the 3 s period. All measurements started at 0 mm. Note that the peak displacements decreased and the time delays increased between (b) and (d). This trend did not change even with the passing of time. Table 2 shows the absolute means ± SD in the

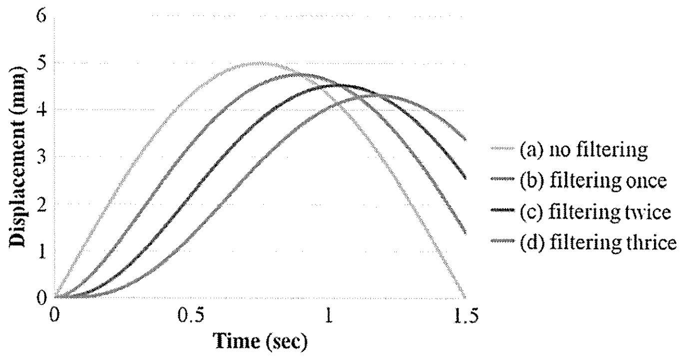

Figure 5. Low-pass filter effects measured for (a) no filtering, (b) filtering once, (c) filtering twice and (d) filtering thrice for the 3 s period. All measurements started at 0 mm.

Table 1. Mean accelerations ± SD and variation coefficients.

	Mean acceleration ± SD (g m^{-2})	Variation coefficient
(a) No filtering	−0.0008 ± 0.0148	−7.52
(b) Filtering once	0.0017 ± 0.0034	2.07
(c) Filtering twice	0.0018 ± 0.0023	1.37
(d) Filtering thrice	0.0022 ± 0.0019	0.96

Table 2. Absolute means ± SD for 3-, 4- and 5 s periods of the reference waveform.

	(a) No filtering versus	Reference waveform periods (sec)		
		3	4	5
Differences in displacement (mm)	(b) Filtering once	0.25	0.15	0.10
	(c) Filtering twice	0.48	0.29	0.19
	(d) Filtering thrice	0.69	0.42	0.28
Differences in time (sec)	(b) Filtering once	0.15	0.15	0.15
	(c) Filtering twice	0.28	0.30	0.30
	(d) Filtering thrice	0.42	0.43	0.45

displacement and time at 3, 4 and 5 s periods of the reference waveform. Note that differences in displacements increased for all periods between (b) and (d) and that displacement for the 5 s period is the smallest. On the other hand, although differences in time increased for all periods between (b) and (d), they are almost same for all periods. From these results, a low-pass filter was used twice in this study.

Figure 6 shows waveforms which were measured at (i) 3-, (ii) 4- and (iii) 5 s periods of the reference waveform. The graphs show comparisons between waveforms measured with the iPod touch® (blue solid line) and on the reference waveform (red dashed line) for 5 breathing cycles. The difference between the measured waveform and the reference waveform was shown by a green solid line. In figure 6 (i), the waveform measured with the iPod touch® shows rather good agreement with the reference waveform, although there are some differences (more details will be discussed in the following chapter). Compared with figure 6 (i), figure 6 (ii)

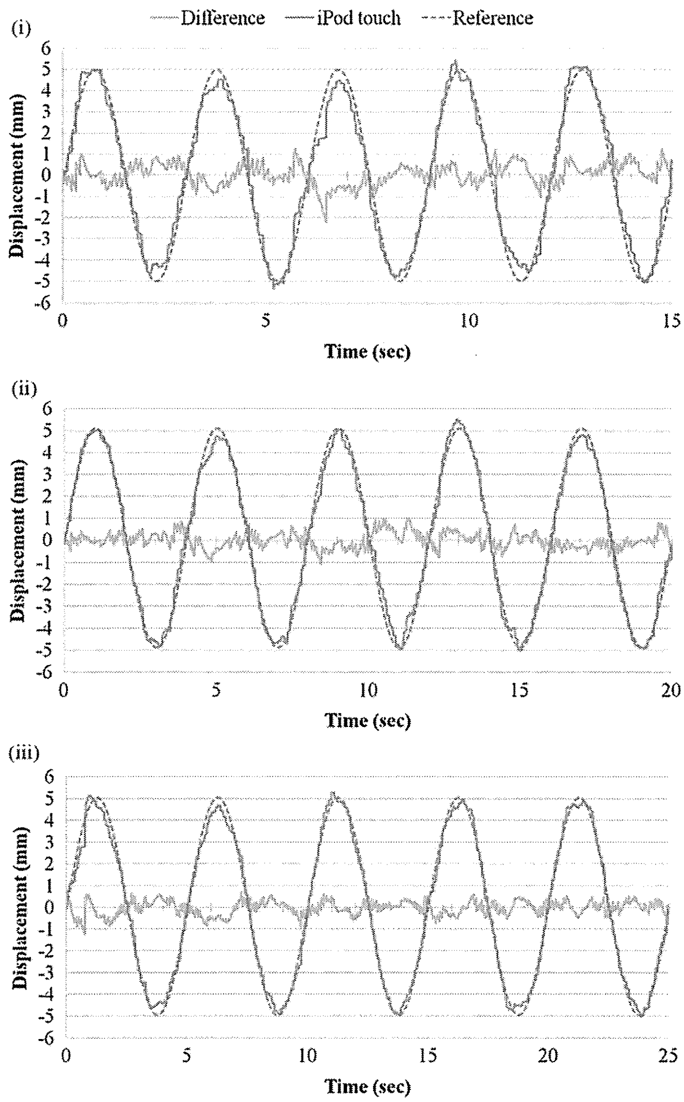

Figure 6. Waveforms measured at (i) 3, (ii) 4 and (iii) 5 s periods of the reference waveform. The graphs show comparisons between the waveforms measured with the iPod touch® (solid line) and on the reference waveform (dashed line). The difference between the measured waveform and the reference waveform is shown by a green solid line.

and (iii) show only minor differences, and good agreement overall with the reference waveform. Table 3 shows the absolute means ± SD of differences in displacement and period between a waveform measured with the iPod touch® and the reference waveform. The absolute means ± SD for displacement during the three periods were (i) 0.45 ± 0.34 mm, (ii) 0.33 ± 0.24 mm and (iii) 0.31 ± 0.23 mm. The value for the displacement of the 3 s period was larger than for the 4 s and 5 s periods. On the other hand, the absolute means ± SD for the three periods were (i) 0.04 ± 0.09 s, (ii) 0.04 ± 0.02 s and (iii) 0.06 ± 0.04 s, showing that values for all periods were almost same.

Table 3. Absolute means ± SD of differences in displacement and period between a waveform measured with the iPod touch® and a reference waveform.

	Reference waveform periods (sec)		
	3	4	5
Displacement (mm)	0.45 ± 0.34	0.33 ± 0.24	0.31 ± 0.23
Period (s)	0.04 ± 0.09	0.04 ± 0.02	0.06 ± 0.04

4. Discussion

4.1. The feasibility of respiratory monitoring

The feasibility of respiratory monitoring using an acceleration sensor (iPod touch®) was evaluated in this study. In theory, a smoother waveform can be obtained by using a low-pass filter many times, but the results in table 1 indicate that the outcomes for filtering two and three times were almost the same. Taking into account weakening of signals and time delays, it was considered optimal to use the low-pass filter twice. The effects of weakened signals and time delays depend on the sampling rate, with a large number of samples diminishing the effects. For this study, the maximum sampling rate was approximately 60 Hz. Figure 6 shows that there were weakened or distorted signals, especially for the 3 s period, for the following three reasons: (1) acceleration noise, (2) effect of the low-pass filter as shown in table 1 and (3) lack of uniformity in the sampling rate. The first reason is illustrated in figure 1, which shows that the low-pass filter could not completely reduce the noise because the acceleration was very noisy. Acceleration was recorded with the system used for this study at a rate of approximately 60 Hz, but it was not uniform and occasionally the recording rate varied between 30 and 120 Hz. This lack of uniformity thus produced distorted signals. The results in tables 2 and 3 make it clear that the 5 s period yielded the most accurate signals, so that the longer time period is suitable for respiratory monitoring.

Christoph *et al* (2005) compared the accuracy of RPM and 3D surface imaging by using a stepping-motor-driven vertical stage which produces a reference waveform. The absolute mean ± SD was 0.03 ± 0.04 mm for the RPM and 0.11 ± 0.15 mm for the 3D surface imaging system (in this case, motion periods were 2–6 s and amplitudes were 1–5 mm). On the other hand, the corresponding value obtained in our study was 0.36 ± 0.27 mm (calculated from table 3 as the average of the 3–5 s periods). However, even though the RPM and 3D surface imaging systems were more accurate than the iPod touch®, our study system is sufficiently accurate for performing respiratory monitoring and has advantages which the other two do not have (these will be discussed in section 4.2).

Our system suffered from time delays because the low-pass filter was used twice to reduce noise of acceleration measurements (figure 5 and table 2). While a smoother respiratory signal can be obtained by adjusting the filter, this would further increase time delays in the respiratory signal. Time delays could not be measured in real time because it was difficult to start measuring the iPod touch® waveform and QUASAR™ waveform at the same time. However, with perform respiratory coaching any difficulties can be overcome, so that the system used in this study is sufficiently accurate for improving the reproducibility of respiratory motion.

4.2. Advantages of using the iPod touch®

There are three advantages to using the iPod touch®. First, a portable consumer electronic device (CED) such as the iPod touch® was used in this study. The accelerometer in the CED is

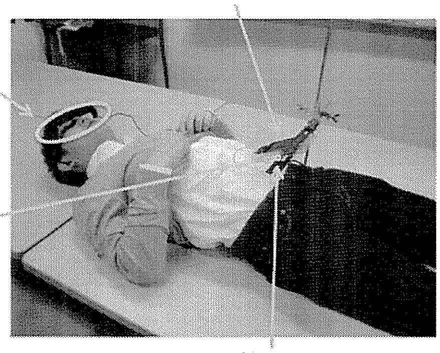

Figure 7. Example of the setup for respiratory training using the iPod touch® and a head-mounted display. The iPod touch® is used for respiratory monitoring and visualization of the respiratory signal. A head-mounted display is used to display the iPod touch® screen to a patient (biofeedback). The support bar and the acrylic plate are affixed by a hinge brace.

quite poor, and even inexpensive devices found in electronic fields are more suitable. Rather, the novelty is the fact that an ordinary CED such as the iPod touch® can be repurposed in this way.

Second, signal detection and signal processing are possible without an external PC because an acceleration sensor is built into the iPod touch® and the required application could be installed with the iPhone software development kit, which provides a variety of original applications. The software was written in Objective-C and special or expensive equipment is not needed because the system used in this study was developed with the iPod touch® only.

Third, the respiratory waveform could be output by connecting the iPod touch® with a head-mounted display. For monitoring the respiratory motion and for respiratory coaching, special or expensive equipment is currently needed. For example, for the RPM system, a charge-coupled device (CCD) camera, a reflective marker, and a computer for processing the data and an output imaging device (for example, a visual screen or a head-mounted display) are needed. In the actual setup, the computer tracks the reflective marker placed on a patient's abdomen by means of a CCD camera and transforms the movement of the reflective marker into a waveform. The output imaging device then shows the patient's respiratory motion. Several investigators have developed devices for respiratory training and reported on their use (Zhang *et al* 2003, George *et al* 2006, Kini *et al* 2002). However, some of these devices require special or expensive equipment and the area for respiratory coaching is limited. Using the iPod touch® and a head-mounted display may make respiratory coaching easier to perform. Figure 7 shows an example of a setup for respiratory coaching using the iPod touch® and a head-mounted display, in which the iPod touch®, which is placed on the acrylic plate, is used for respiratory monitoring and visualization of the respiratory signal while the patient can see the iPod touch® screen on a head-mounted display (biofeedback). The support bar and the acrylic plate were affixed by a hinge brace, so these positions did not change. On the other hand, the contact point of the acrylic plate against the abdomen was not affixed. A smooth plate was placed on the abdominal surface. The acrylic plate could slide on the smooth plate. Because this respiratory coaching tool is portable and inexpensive, patients can perform respiratory coaching at home or in the hospital ward and access training more frequently.

5. Conclusion

In this study, the feasibility of a respiratory monitoring system using an acceleration sensor was evaluated. The accuracy of respiratory monitoring with the acceleration sensor was satisfactory, judging from the results for the absolute means $\pm$ SD. Using the iPod touch® for respiratory monitoring does not require any special equipment and makes respiratory monitoring easier. For these reasons, this system is viable for respiratory monitoring.

Acknowledgments

This study was supported by a Grant-In-Aid for Scientific Research from the Japan Society for the Promotion of Sciences (no. 23390300).

References

Bruce E N 1996 Temporal variations in the pattern of breathing *J. Appl. Physiol.* **80** 1079–87

Christoph B, Metheany K G, Karen D and Chen G T Y 2005 A phantom evaluation of a stereo-vision surface imaging system for radiotherapy patient setup *Med. Phys.* **32** 2753–62

George R, Chung T D, Vedam S S, Ramakrishnan V, Mohan R, Weiss E and Keall P 2006 Audio-visual biofeedback for respiratory-gated radiotherapy: impact of audio instruction and audio-visual biofeedback on respiratory gated radiotherapy *Int. J. Radiat. Oncol. Biol. Phys.* **65** 924–33

Jonathan K H, Daniel B P, Byong Y Y and Cedric X Y 2008 On the sources of drift in a turbine-based spirometer *Phys. Med. Biol.* **53** 4269–83

Kini V R, Subrahmanya S V, Keall P J, Sumukh P, Clayton C and Radhe M 2002 Patient training in respiratory-gated radiotherapy *Med. Dosim.* **28** 7–11

Laura I, Sonia G, Mary A, Catheryn Y and Steve B J 2009 Using surface imaging and visual coaching to improve the reproducibility and stability of deep-inspiration breath hold for left-breast-cancer radiotherapy *Phys. Med. Biol.* **54** 6853–68

Masselli G M P, Silvestri S, Ramella S and Trodella L 2009 Design and evaluation of a methodology to perform personalized visual biofeedback for reducing respiratory amplitude in radiation treatment *Med. Phys.* **36** 1467–72

Neicu T, Berbeco R, Wolfgang J and Jiang S B 2006 Synchronized moving aperture radiation therapy SMART: improvement of breathing pattern reproducibility using respiratory coaching *Phys. Med. Biol.* **51** 617–36

Ozhasoglu C and Murphy M J 2002 Issues in respiratory motion compensation during external-beam radiotherapy *Int. J. Radiat. Oncol. Biol. Phys.* **52** 1389–99

Shimizu S, Shirato H, Ogura S, Akira-Dosaka H, Kitamura K, Nishioka T, Kagei K, Nishimura M and Miyasaka K 2001 Detection of lung tumor movement in real-time tumor-tracking radiotherapy *Int. J. Radiat. Oncol. Biol. Phys.* **51** 304–10

Shirato H *et al* 2000 Physical aspects of a linear accelerator synchronized with real-time tumor tracking system *Int. J. Radiat. Oncol. Biol. Phys.* **48** 1187–95

Tobin M J, Jenouri G, Birch S J, Gazeroglu H B and Sackner M A 1983a Breathing patterns: 1. Normal subjects *Chest* **84** 202–5

Tobin M J, Jenouri G, Birch S J, Gazeroglu H B and Sackner M A 1983b Breathing patterns: 2. Diseased subjects *Chest* **84** 286–94

Vedam S S, Keall P J, Kini V R and Mohan R 2001 Determining parameters for respiration-gated radiotherapy *Med. Phys.* **28** 2139–46

Vedam S S, Kini V R, Keall P J, Ramakrishnan V, Mostafavi H and Mohan R 2003 Quantifying the predictability of diaphragm motion during respiration with a noninvasive external marker *Med. Phys.* **30** 505–13

Zhang T, Keller H, O'Brien M J, Mackie T R and Paliwal B 2003 Application of the spirometer in respiratory gated radiotherapy *Med. Phys.* **30** 3165–71