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Pendelluft is not the major contributor to respiratory insufficiency in dogs with flail chest: a mathematical analysis

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Abstract: "Pendelluft", or out-of-phase movement of the airway gas between the intact and flail-chest-side lungs has long been believed to be the major contributor to respiratory dysfunction in patients with flail chest. However, conflicting findings have also been reported mainly from animal studies. The aim of this study was to provide a mathematical projection on this classical problem. We measured respiratory impedance (Z_{RS}) of dogs with flail chest using a pseudorandom forced oscillation method. A mathematical model implementing flail chest was fitted to Z_{RS}. The fitted results were used in simulating the mechanical behavior of a respiratory system with flail chest during spontaneous breathing. Our results suggest that the paradoxical movement of breathing between the flail segment and the intact chest wall does not create substantial pendelluft and that alveolar hypoventilation is created by the wasting movement of the flail segment which interferes with effective thoracic expansion.

Key words: Flail chest, Pendelluft, Impedance, Mathematical model

Introduction

Double fractures of multiple contiguous ribs produce flail chest which is characterized by mechanical instability typically manifested by paradoxical movement of the chest wall [1]. We often encounter flail chest after thoracic surgery or trauma. It has long been said that flail chest produces respiratory insufficiency expressed by the term "pendelluft", i.e., to-and-fro air flow between the lungs of the intact and flail chest sides during spontaneous breathing [1]. However, conflicting results have been reported where pendelluft does not necessarily play a major role in respiratory failure [2,3]. Therefore, we desired to explain the differences among those studies by introducing mathematical models into respiratory mechanics in an experimentally produced flail chest.

The study consists of two parts. First, we assessed the effect of flail chest on respiratory mechanics by measuring the impedance of the respiratory system in anesthetized and paralyzed dogs. The implication of impedance in the respiratory mechanics is described in Appendix A. Inverse modeling was performed with a five-compartment model in which the effect of flail chest on respiratory mechanics was interpreted by the changes in the model parameters. Second, employing a simplified version of the model, we used a computer to simulate the flail chest movement during spontaneous breathing and quantified the extent of pendelluft.

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Materials and methods

Experimental

Six mongrel dogs weighing 17–23 kg were anesthetized with pentobarbital sodium 25 mg·kg^{-1}. Anesthesia was maintained with hourly doses of 30–40 mg pentobarbital. The dogs were tracheostomized, and a tracheal cannula was inserted. Airway flow ($\dot{V}$) was measured with a Fleisch #2 pneumotachograph connected to a differential piezoresistive transducer (Microswitch 163PC01D36, Honeywell, Tokyo, Japan). The pneumotachograph was, in turn, connected to a three-way valve which made it possible to switch between a mechanical ventilator (EVA-900, Aika, Tokyo, Japan) and a volume oscillator described elsewhere [4]. Tracheal pressure (Ptr) was measured by a piezoresistive pressure transducer (ICS 13 002G, ICSensors, Milpitas, CA, USA) at a side port of the tracheal cannula. The dogs were paralyzed with pancuronium bromide (4 mg·h^{-1}) and mechanically ventilated with a tidal volume of 12 ml·kg^{-1} with air supplemented by oxygen. End-tidal carbon dioxide tension was monitored with a respiratory gas monitor (Capnox, Nippon Colin, Tokyo, Japan) and maintained at 35–45 mmHg.

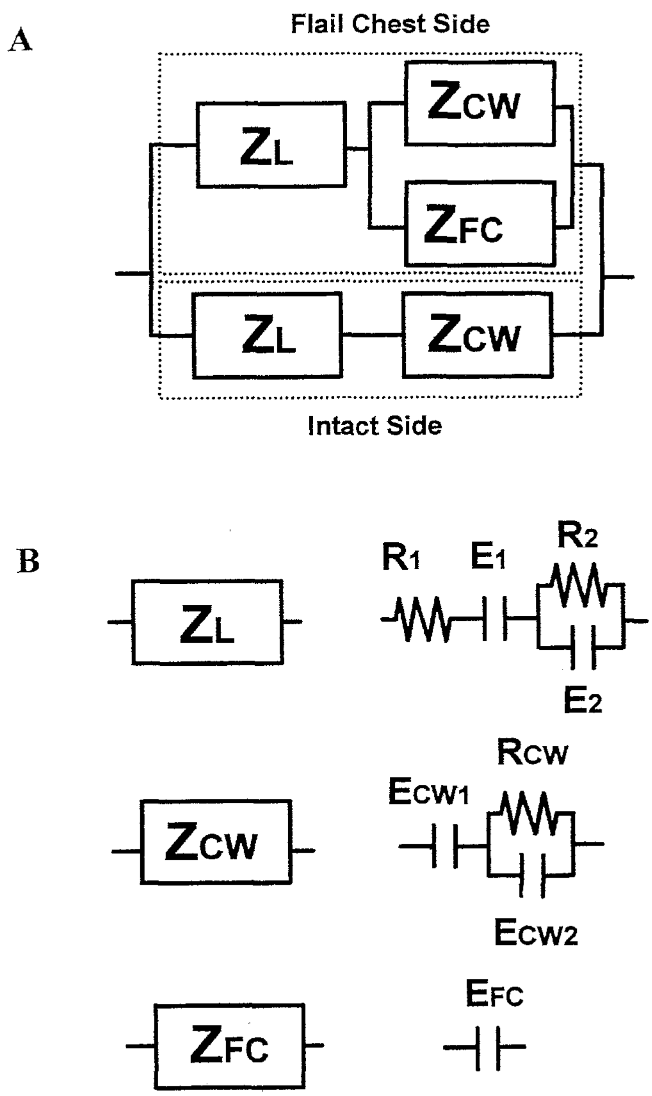

Fig. 1. **A** The model fitted to respiratory impedance (ZRS) measured in anesthetized and paralyzed dogs. The model consists of two subsystems representing intact and flail-chest sides which are connected in parallel. The intact subsystem is comprised of lung impedance (*ZL*) connected in series with chest wall impedance (*Zcw*). On the flail-chest side, an additional impedance (*Zfc*) representing flail-chest mechanics is connected. We assumed that both ZL and Zcw were identical in both subsystems. **B** The electrical analogues of the compartmental impedances employed in the model. The model for ZL (from [6]) to interpret viscoelastic characteristics of lung tissue. *R1* denotes airway resistance. The viscoelastic behavior of lung tissue is represented by the combination of a resistance (*R2*) and two elastances (*E1* and *E2*). Zcw has the same structure as ZL except the lack in the resistance corresponding to *R1* in ZL. ZFC has only one element of elastance (*EFC*)

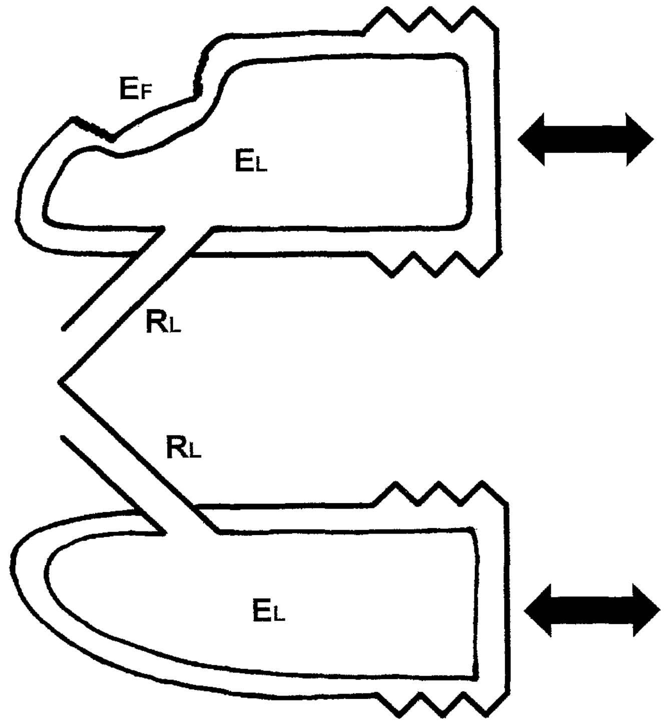

Fig. 2. Model employed in simulating the effect of flail chest on respiratory mechanics during spontaneous breathing. The model comprises two respiratory subsystems, one representing the intact and the other the flail-chest sides, which are connected in parallel. The *lower half* in the figure represents the intact subsystem which consists of lung with a resistance (*RL*) and an elastance (*EL*) enclosed by a chest wall. The chest wall has a bellows producing the generating force for spontaneous breathing as the mechanical analogue of a hemidiaphragm. The *upper half* expresses the subsystem suffering from a flail chest. The lung has exactly the same structure as that in the intact side with the identical values of both RL and EL. The flail chest is expressed by an elastance (*EFC*) on the chest wall. The elastance, EFC, is connected mechanically in parallel with the lung. In the simulation, the bellows of both subsystems oscillate completely in phase and with the same stroke volume

Mechanical impedance of the respiratory system (ZRS) was measured in a frequency range between 0.15 and 6.65 Hz by a pseudorandom forced oscillation method as described previously [4,5]. Briefly, mechanical ventilation was stopped at end expiration and the three-way valve was switched to the volume oscillator. A pseudorandom signal sent from a computer drove the volume oscillator which, in turn, oscillated the respiratory system. The measurement was repeated five times with a 2-min interval of mechanical ventilation between each run, and five sets of measured signals were ensemble averaged. ZRS was computed with the averaged signals using an auto- and cross-correlation method [4] [(zero end-expiratory pressure (ZEEP) control condition]. Following the measurements, the dogs were ventilated with a positive end-expiratory pressure (PEEP) of 5 cmH_2O and ZRS measurements were repeated (PEEP control).

After the control measurements we constructed a flail chest by producing double fractures at each of the 4th to 7th ribs of the right chest wall. A complete fracture was secured by cutting an entire piece of the rib away at the fractured site. The length of the fractured ribs varied from 5 to 15 cm according to the size of the rib. We confirmed that the flail segment of the chest wall was completely separated from the intact portion. We created small holes on the parietal pleura communicating between the atmosphere and the pleural space on both sides of the chest wall. We confirmed every 30 min that pneumothorax was not produced since no air was observed to escape from the pleural space through the holes when high airway pressure was applied. Otherwise, the holes were tightly sealed with no air leakage throughout the experiment.

The measurements of ZRS were repeated with zero and 5 cmH_2O end-expiratory pressure, in the same way as the control measurements (ZEEP and PEEP flail chest conditions, respectively). All signals were low-pass filtered at 30 Hz (SPA-3 3-pole Butterworth filter, TechnoService, Chiba, Japan), digitized at 102.4 Hz (DT2801-A analog-to-digital converter, Data Translation, Marlboro, MA, USA) and stored in an AT 286 computer. The data were collected and processed by ANADAT and LABDAT data acquisition and analysis software (RHT-InfoDat, Montreal, Canada).

The study was approved by the Laboratory Animal Center Ethics Committee, School of Medicine, Chiba University.

Model fitting and simulation

We fitted a model to the measured ZRS using a nonlinear least-squares technique described by Sato et al. [4]. The model consists of five compartments as illustrated in Fig. 1. The best-fit model of ZRS and the estimate of the parameter EFC, representing the flail segment elastance, were used in the following simulation model of spontaneous breathing.

We employed a simplified version of the model for simulating respiratory mechanics with flail chest during spontaneous breathing. The model is shown in Fig. 2. The simulated spontaneous breathing was sinusoidal at a frequency of 0.25 Hz (15 breaths·min^{-1}). The contributions from the real (RRS) and imaginary (XRS) parts of ZRS from the best-fit model were converted to a corresponding elastance value at 0.25 Hz and substituted into the spontaneous breathing model as the resistance and the elastance of the lung subsystems. We also chose the EFC value of the best-fit model obtained from the five-compartment modeling as the flail segment elastance (EFC of the spontaneous breathing model). If the excitation of a system is sinusoidal at any given frequency, the

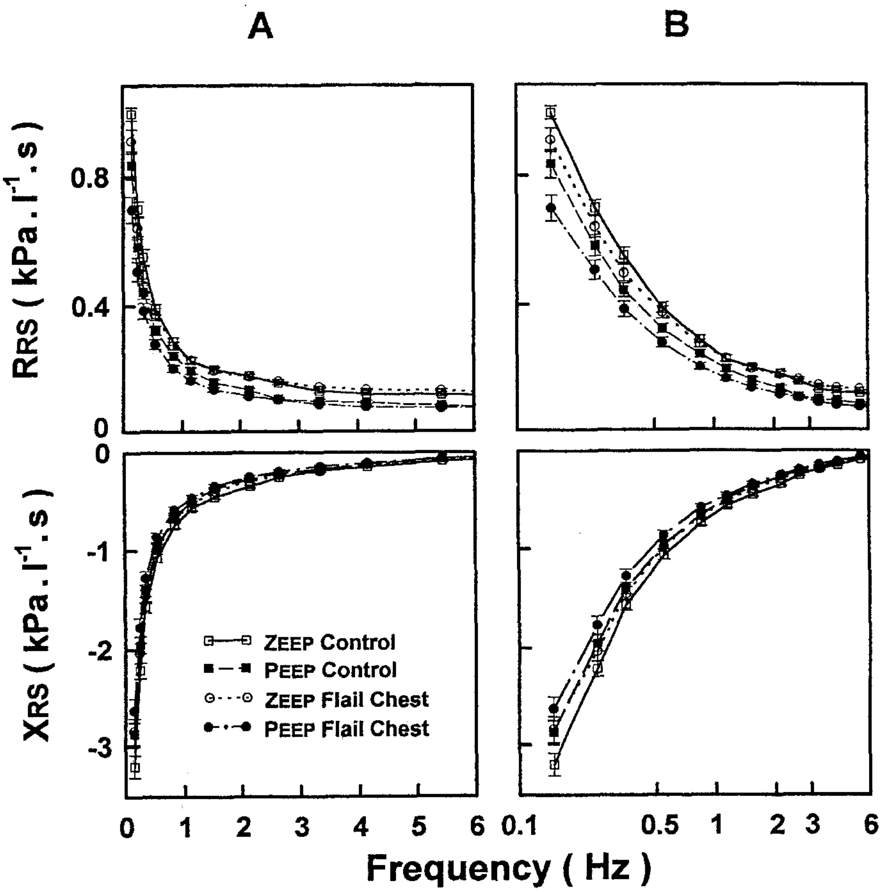

Fig. 3. Respiratory system impedance (ZRS) in control and flail-chest conditions. The data are the averages obtained from six dogs (mean ± SEM). The *upper panel* shows respiratory system resistance (*RRS*, the real part of ZRS) and lower panels the reactance (*XRS*, the imaginary part of ZRS). In control condition *open squares* and *closed squares* denote ZRS measured with ZEEP and PEEP, respectively. In flail-chest condition *open circles* and *closed circles* denote ZRS with ZEEP and PEEP, respectively. In the *left panels*, ZRS are plotted on a linear scale to show the frequency-dependent behavior of ZRS. In the *right panels*, they are plotted on a natural logarithmic scale to show the differences in ZRS between four conditions more clearly, in particular at low frequencies

system can be assumed to behave as a single compartment system, i.e., to have a single time constant. Spontaneous breathing is generated only by the diaphragm, which is represented by the bellows in Fig. 2. The bellows of both sides moved completely in phase and with an identical stroke volume. We examined the extent of pendelluft, i.e., the phase and tidal volume discrepancies between the intact and the flail-side lungs. We employed three EFC values, the EFC value of the best-fit model and values an order of magnitude higher and lower, to examine the effect of the flail segment elastance on the magnitude of pendelluft.

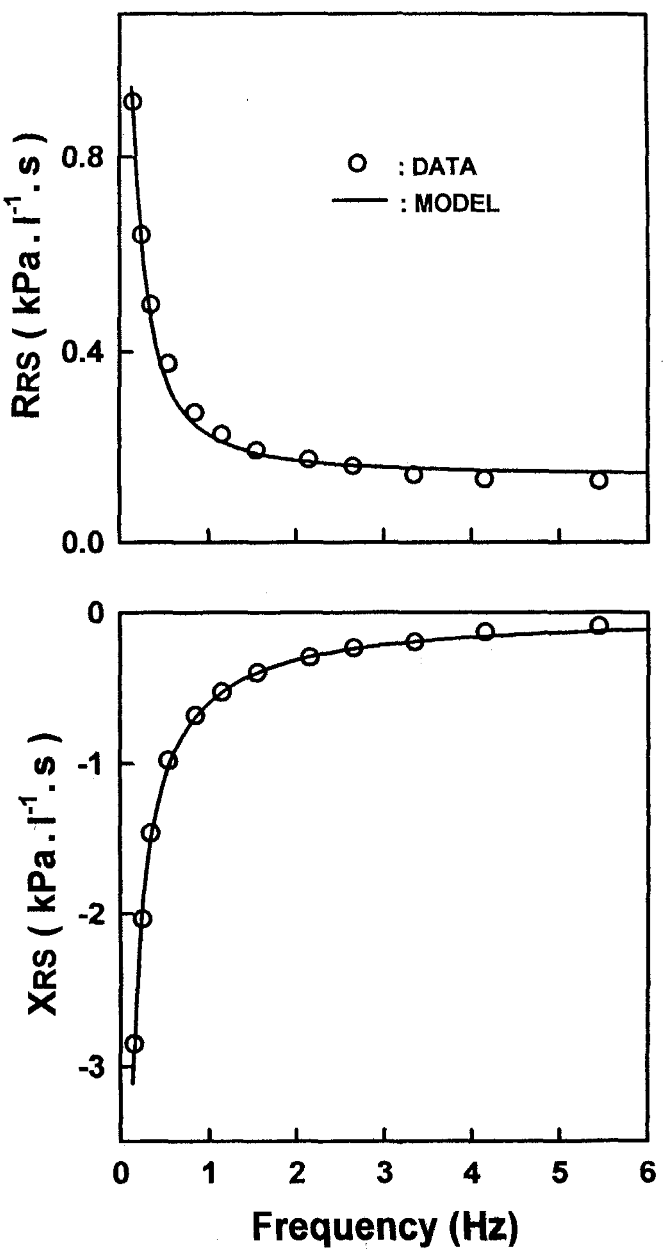

Fig. 4. An example of fitting of the five-compartment model to the measured ZRS in ZEEP flail-chest condition in a representative dog. *Open circles* denote the measured ZRS and *solid lines* the best-fit model

Results

Figure 3 shows the measured ZRS in the four conditions. The respiratory system resistance (RRS, the real part of ZRS) decreased hyperbolically with frequency in the control measurements either with or without PEEP (Fig. 3A). The flail chest did not change the hyperbolic shape, indicating that uneven ventilation was not substantial during forced oscillation [4,5]. The effect of the flail chest on the magnitude of ZRS was so complicated that it was difficult to compare the values of ZRS between the conditions. We therefore introduced another model fit as described in Appendix B to examine the effects of flail chest and PEEP on ZRS statistically. The modeling test suggested that: (1) the flail chest did not change RRS; (2) PEEP reduced RRS both in the control and flail chest conditions; and (3) both the flail chest and PEEP decreased XRS.

Figure 4 illustrates an example of our five-compartment model best-fitted to the ZRS in a representative dog. The model exhibited a superior fitting in all conditions, suggesting the model was able to characterize the mechanical effect of flail chest on the respiratory system, although the model parameters were not necessarily well determined.

Figure 5 shows the results of simulating flail chest movement during spontaneous breathing. A decrease in EFC produced a significant decrease in tidal volume of the flail-chest side (Fig. 5A), though the phase discrepancy between the intact and the flail-chest sides was very small with any EFC (Fig. 5B).

Discussion

Our findings in ZRS measurement are that: (1) RRS exhibited hyperbolic frequency-dependent behavior whether flail chest existed or not; (2) flail chest did not produce an appreciable change in RRS; (3) PEEP decreased RRS in both control and flail chest conditions; and (4) both flail chest and PEEP reduced XRS.

When ventilation nonhomogeneity becomes substantial, lung resistance alters its hyperbolic frequency dependence to exponential dependence [4,5], which should be accompanied by alteration in the shape of RRS by the same extent. This indicates that, during forced oscillation, flail chest does not produce significant uneven ventilation, i.e., to-and-fro movement of airway gas between intact and flail chest side lungs.

The effects of flail chest on respiratory mechanics may be twofold: one due to the reduction in lung vol-

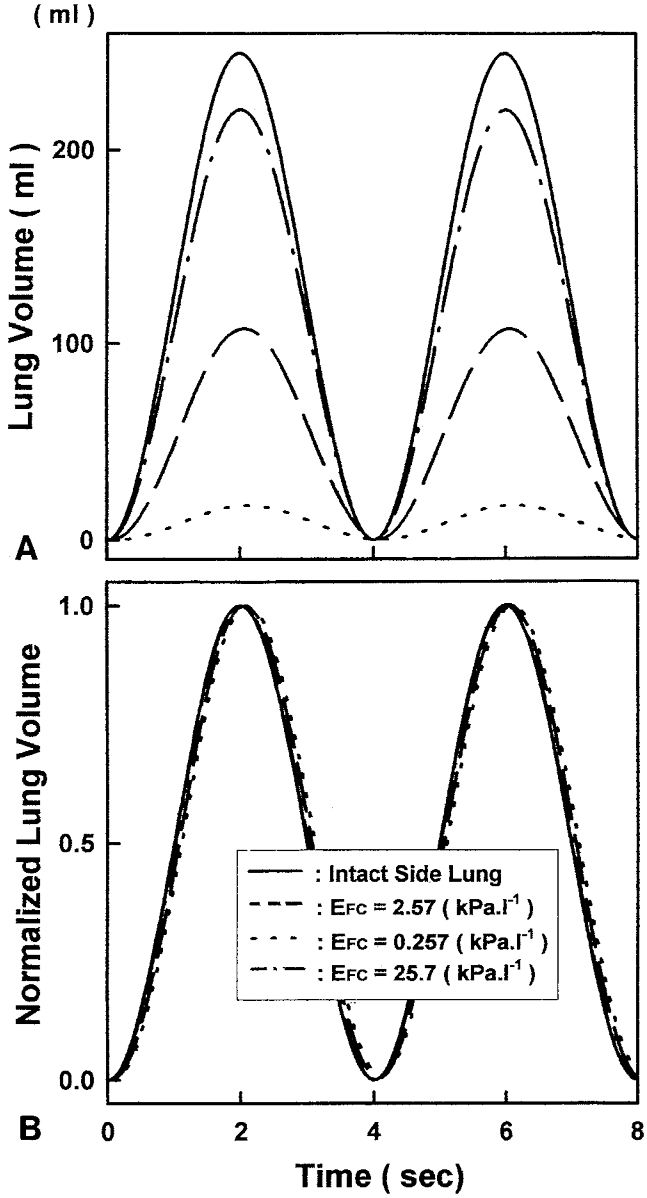

Fig. 5. A The result of simulating the effect of flail chest on a sinusoidal spontaneous breathing at 0.25 Hz. The tidal displacements of hemidiaphragms of both sides were 250 ml and they were completely in phase. As expected, the effect of EFC is obvious on lung volume variation of the flail-chest side. **B** The normalized lung volume variations in the simulation shown to examine the phase difference in lung volume variations between the intact and the flail chest side lungs. Minimal phase differences suggest that pendelluft is not appreciable in the simulation

ume below functional residual capacity (FRC) and the other due to the presence of a floppy chest wall segment. Barnas et al. [7] demonstrated that in awake sitting humans both lung resistance and elastance increased with decreasing lung volume below FRC. By contrast, Dechman et al. [8] reported that in open-chest supine dogs, lung elastance increased with decreasing lung volume whereas lung resistance did not show a consistent change unless FRC was extremely low. They also showed increasing contribution of airway resistance to lung resistance as lung volume decreased from FRC. In our measurements of ZRS, RRS did not show a significant change. The modeling results described in Appendix B suggested that the resistances of neither respiratory tissue represented by the parameter B nor the airway by RAW were changed by flail chest. These indicate that the reduction in lung volume by the flail chest, if any, was not large enough to produce appreciable changes in respiratory mechanics. The decrease in XRS, which was reflected in the decrease in parameter A of the model in Appendix B, was probably due to the discontinuity between the flail segment and the intact chest wall. The flail segment works as a highly compliant element in a respiratory system.

The reduction in ZRS by PEEP might be due to lung volume expansion above FRC level and bimodal, i.e., the decreases in both airway and tissue impedances. In the anesthetized supine position, in which FRC is lower than other postures [9,10], airway derecruitment and atelectasis are characteristic features as well as the reduction of airway caliber [7,8,11]. PEEP may decrease airway resistance by increasing the airway caliber. PEEP might also reduce nonlinear mechanical phenomena such as pop-on and -off of closed airways, and formations of menisci and atelectasis, all of which act as unpredictable resistive and elastic elements in the respiratory tissue [8,11–13]. Our results are consistent with the result of Barnas et al. [7] suggesting that a moderate amount of PEEP would decrease lung resistance and elastance around the physiological FRC level.

The model fitted to ZRS data had five compartments with eight parameters in total (Fig. 1). We believe each compartment to have the simplest and most reasonable representation of the physiological mechanical conditions. It was necessary for each compartment in the intact side to be identical to its counterpart in the flail chest side in order to keep the number of mechanical elements as small as possible for computational simplicity. We also assumed the flail segment to be represented by a single linear elastance. However, the model was still complicated, and some of the model parameters were often overdetermined although the fitting was always extremely good (Fig. 4). In a mathematical sense, this problem was in part because the RRS data retained its simple hyperbolic shape even in the presence of flail chest. This, in turn, was indirect evidence that appreciable pendelluft did not occur in flail chest and justified our model structure.

In our simulation of respiratory mechanics with flail chest during spontaneous breathing, the striking part is that the phase discrepancy in ventilatory volume between the flail chest and the intact sides was extremely small regardless of the wide variation of EFC (Fig. 5B). Pendelluft, the volume of to-and-fro gas movement between the lungs, amounted to less than 2% of total ventilation. We attributed a single linear elastance to the flail segment which, in reality, should be highly nonlinear since its oscillatory movement was limited, i.e., not linearly proportional to the applied pressure. However, our simulation with a two order of magnitude range of EFC should cover the nonlinear effect, suggesting that the phase difference between the sides may remain small even with a nonlinear EFC. The mediastinal septum separating the lungs is not rigid, therefore, ventilation of each lung is, to some extent, dependent on the other lung due to the pleural pressure gradient [3,12]. However, the mediastinum may function mainly as an elastic element so that it would not create appreciable further phase difference between the ventilation of the lungs.

Our simulation suggest that flail chest produces hypoventilation inversely proportional to the elastance of the flail segment. That is, a part of the tidal movement of the intact chest wall is wasted by movement of the flail segment in the opposite phase.

In summary, we measured ZRS in dogs with flail chest. A mathematical model representing a respiratory system with flail chest was fitted to ZRS. Using the fitted results, we simulated a model of a respiratory system with flail chest during spontaneous breathing. Both ZRS measurements and the simulation suggested that pendelluft is not significant and that hypoventilation produced by the opposite directional movement of the flail segment against the intact chest wall is the major cause of respiratory insufficiency in flail chest.

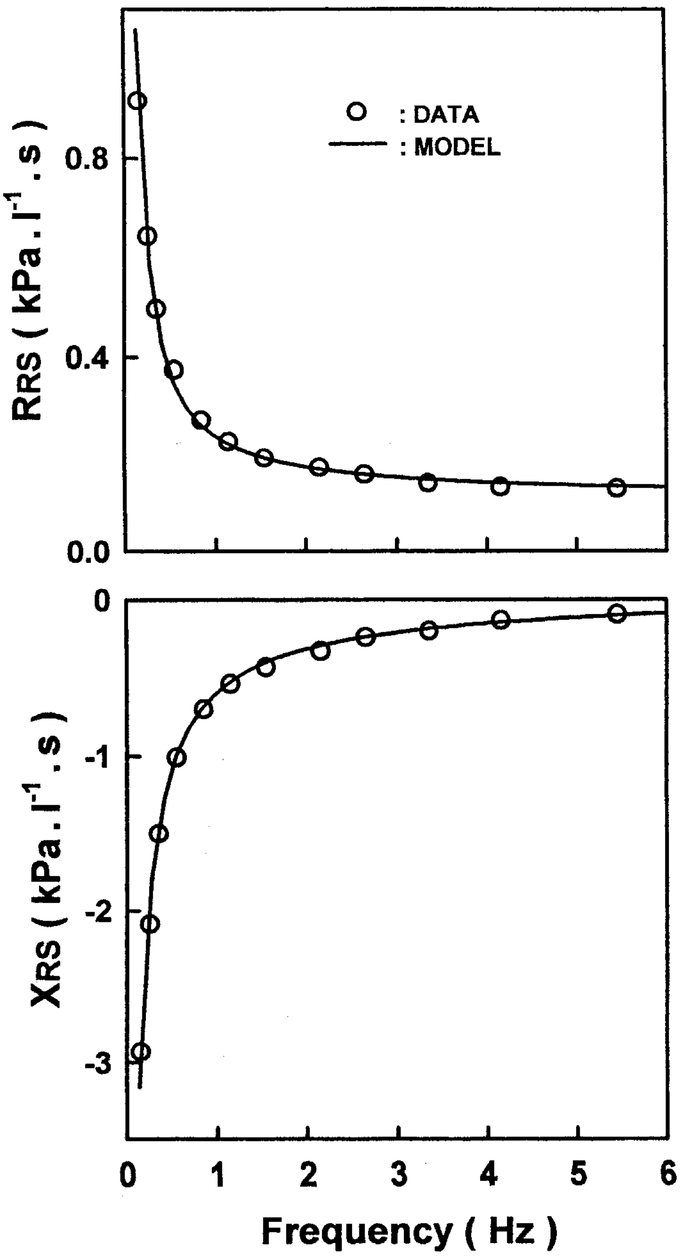

Fig. B1. Example of the four-parameter model best-fitted to ZRS data in ZEEP flail-chest condition in a representative dog. *Open circles* and *solid lines* denote the data and the model, respectively

Appendix A

In the studies of respiratory mechanics, two of the most familiar parameters are resistance (R) and elastance (E). Compliance is the inverse of E. Whenever we refer to these parameters, we implicitly assume a mechanical model a priori, which is formulated by the following differential equation,

$$\mathrm{P_{AW}} = \mathrm{R}\dot{\mathrm{V}} + \mathrm{EV} \quad \text{(A1)}$$

where PAW, $\dot{V}$, and V denote airway pressure, airway flow and ventilation volume, respectively.

If we see a respiratory system as a linear mechanical system, it possesses its own system function. The system function (also called transfer function) does not require any a priori model. If we know the system function, we can accurately predict the response to whatever perturbation is input to the system. In a respiratory system, the input and output are flow ($\dot{V}$) and pressure (PAW) at the airway opening, respectively, and the system function is called impedance (ZRS). ZRS is a complex function (a composite function of real and imaginary parts) of frequency. The real and imaginary parts are referred to as resistance and reactance, respectively. ZRS of the respiratory system formulated by Eq. A1 is,

$$\mathrm{Z_{RS}}(\omega) = \mathrm{R} - i\mathrm{E}/\omega \quad \text{(A2)}$$

where ω and i are angular frequency (2πf, f: frequency) and the square root of minus one, respectively. As

Table B1. Values of parameters obtained by fitting the additional four-parameter model to ZRS

	Control		Flail chest	
	ZEEP	PEEP	ZEEP	PEEP
RAW ($\times 10^{-1}$kPa·l^{-1}·s)	1.02 ± 0.08	0.75 ± 0.26*	1.10 ± 0.14	0.69 ± 0.05*
IAW ($\times 10^{-3}$kPa·l^{-1}·s^2)	1.39 ± 0.31	1.74 ± 0.22	1.47 ± 0.34	1.55 ± 0.38
A (kPa·l^{-1})	2.75 ± 0.22	2.52 ± 0.22*	2.49 ± 0.26#	2.30 ± 0.24*,#
B (kPa·l^{-1})	1.32 ± 0.09	0.98 ± 0.22*	1.19 ± 0.15	0.93 ± 0.14*

Values are means ± SD from six dogs.
* $P < 0.05$ between ZEEP and PEEP in each condition of control or flail chest; # $P < 0.05$ between control and flail chest at either ZEEP or PEEP.
ZEEP, zero end-expiratory pressure; PEEP, positive end-expiratory pressure; RAW, airway resistance; IAW, airway inertance; A, elastance of respiratory tissue; B, resistance of respiratory tissue.

shown in Eq. A2, the fundamental element in the reactance is elastance (E).

Appendix B

ZRS is a function of frequency, which prevents us from assessing the effect of disturbances to a respiratory system simply by comparing the value of ZRS before and after the disturbance. Therefore, we resorted to additional mathematical modeling. The measured ZRS were fitted to a model which provided the parameter estimates. The effects of flail chest and PEEP were statistically tested by comparing changes in the model parameters. The model employed consisted of a single conduit connected in series with an alveolar region surrounded by viscoelastic respiratory tissue, which is formulated as,

$$Z_{RS}(\omega) = R_{AW} + B\pi/4.6\omega + i\omega I_{AW} - i(A + 0.25B + B\,log\,\omega)/\omega$$

where RAW and IAW denote airway resistance and inertance, and A and B respiratory tissue elastance and resistance, respectively.

The model is widely accepted to characterize the respiratory mechanics in normal conditions [3,4], though it does not consider uneven ventilation. We used this model since the flail chest was thought not to produce appreciable ventilation nonhomogeneity, as discussed in the Results and Discussion. ZRS measured in our frequency range reflects the mechanical properties of both airways and the respiratory tissue (lung parenchyma and chest wall). RRS at low frequencies, largely the frequency-dependent segment, reflects mainly the tissue resistive property, which is represented by the parameter B, while the frequency-independent asymptotic segment at high frequencies indicates airway resistance which is expressed by RAW [3,4]. XRS is characterized by respiratory tissue mechanical constituents, i.e., its resistive (B) and elastic (A) properties. The inertance of airway gas (IAW) is also included in XRS though its contribution is very low in the physiological frequency range.

An example of the model best-fitted to the ZRS data in the ZEEP flail chest condition is shown in Fig. B1. The fitting was extremely good over a range of conditions.

The results are presented in Table B1, and the statistical difference was tested by a repeated measures ANOVA with Fisher's test and considered significant when $P < 0.05$. The results showed: (1) that the flail chest did not change the values of RRS, RAW, and B but PEEP reduced both RAW and B; (2) that both flail chest and PEEP reduced the tissue elastance parameter A which affects XRS.

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