Mechanics, nonlinearity, and failure strength of lung tissue in a mouse model of emphysema: possible role of collagen remodeling

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tory lung volume was performed with a computer-controlled ventilator without the application of positive end-expiratory pressure (PEEP) (21). Dynamic respiratory mechanics were measured at four different PEEP levels (0, 3, 6, and 9 cmH₂O) in the closed-chest condition and assessed by measuring impedance data during forced oscillations. To standardize volume history, each measurement was preceded by two consecutive inflations of the lungs to total lung capacity.

**Impedance measurements.** Impedance data collection was made by interrupting mechanical ventilation for 6 s by use of the optimal ventilation waveform (OVW), which is a broadband waveform containing energy from 0.5 to 15 Hz as described previously (27). The frequencies in the OVW are selected according to a nonsum-nondifference criterion, which eliminates harmonic distortion and minimizes cross talk among the frequencies present in the input flow waveform and hence provides smooth estimates of the input impedance (43). The volumes delivered are similar to normal spontaneous tidal volume values; hence, the method provides information on the mechanical properties during conditions mimicking breathing. In our experiment, we matched the peak-to-peak OVW amplitude to the tidal volume delivered by the mechanical ventilator. The ventilator displacement and cylinder pressure signals were low-pass filtered at 30 Hz and sampled at 256 Hz. With the use of Fourier analysis, impedance spectra were calculated on overlapping blocks of pressure, and flow data were calculated as the ratio of the cross-power spectrum of pressure and flow and the autopower spectrum of flow. The forced-oscillatory system was calibrated by measuring the input impedance of known analogs, including tubes and bottles with known impedances. The frequency response of the system was obtained, and the measured impedance spectra were off-line corrected for any phase difference between pressure and flow. Additionally, the flow-dependent impedance of the tracheal cannula was characterized separately and removed from the respiratory impedance of the mice (20).

**Mathematical modeling.** Data were analyzed with a model that allows for specific alterations in the lung tissue similar to what might be expected in emphysema. Specifically, we applied a heterogeneous tissue elastance model of the lung (20). Briefly, in this model, we represented the airway tree by a set of airway pathways arranged in parallel, where each compartment is composed of an airway resistance and tissue elastance model of the lung (20). Briefly, in this model, we matched the peak-to-peak OVW amplitude to the tidal volume delivered by the mechanical ventilator. The ventilator displacement and cylinder pressure signals were low-pass filtered at 30 Hz and sampled at 256 Hz. With the use of Fourier analysis, impedance spectra were calculated on overlapping blocks of pressure, and flow data were calculated as the ratio of the cross-power spectrum of pressure and flow and the autopower spectrum of flow. The forced-oscillatory system was calibrated by measuring the input impedance of known analogs, including tubes and bottles with known impedances. The frequency response of the system was obtained, and the measured impedance spectra were off-line corrected for any phase difference between pressure and flow. Additionally, the flow-dependent impedance of the tracheal cannula was characterized separately and removed from the respiratory impedance of the mice (20).

By minimizing the root mean square difference between the model and the data (6), the five parameters (Raw, Iaw, η, Hmin, and Hmax) were determined. The mean H value (Hmean) was estimated as the expected value of the distribution function and was calculated from the estimates of Hmin and Hmax as

\[
H_{\text{mean}} = \frac{H_{\text{max}} - H_{\text{min}}}{F}
\]

where Hmean, Hmax, and Hmin are given by F = ln(Hmax/Hmin), Ztimin (ω) = (η – j)Hmin/ω0, and Ztimax (ω) = (η – j)Hmax/ω0. The tissue damping, G, was calculated as G = ηHmean.

**Dynamic nonlinearity of respiratory mechanics.** When applying a broadband input, one way to probe system nonlinearity is to measure how the elastic and viscous moduli depend on the amplitude of the input. Another way of quantifying nonlinearity is via the so-called harmonic distortion index, which estimates the amount of both harmonic distortion and cross talk in the output signal resulting from system nonlinearities (42). For a broadband input, the coefficient of harmonic distortion (k_d) is defined as

\[
k_d = \sqrt{P_{\text{tot}}/P_{\text{NI}}} \times 100\%\]

where Ptot is the total power in the output and PNI is the output power due to system nonlinearities only, i.e., the power at noninput frequencies. Because only nonlinearities and noise can produce output energy at noninput frequencies, the values of k_d were also corrected for nonzero energy at noninput frequencies (54). The advantage of using k_d is that it can be calculated from a single impedance measurement, whereas the traditional method of characterizing nonlinearity requires the measurement of the moduli at several distinct amplitudes. The k_d in a linear system is zero. In a nonlinear system driven by sinusoids, the k_d measures the distortion and cross talk due to system nonlinearities.

**Measurements of mechanical failure of lung tissues.** To assess the strength of the alveolar wall in the parenchyma, failure tests of lung tissue strips were carried out. Similar to the measurements of respiratory function, an additional dose of pentobarbital sodium (70 mg/kg) was injected intraperitoneally to each animal (n = 6 in each group); the thorax was then opened, and the animals were exsanguinated by severing the inferior vena cava. The heart, lungs, and trachea were carefully dissected en bloc and rinsed in PBS (Sigma). The experimental setup was described previously (53). Parenchymal tissue strips having dimensions of 2.0–3.0 mm × 0.7–1.0 mm × 0.7–1.0 mm in length, width, and thickness, respectively, were carefully prepared from each lung, and the pleura were removed with the use of a razor. Each end of the tissue strip was fixed by cyanoacrylate glue to small metal plates attached to straight steel wires. The assembly was placed in a horizontal tissue bath filled with PBS at room temperature, with one wire attached to a computer-controlled lever arm containing a force transducer (model 300B, Aurora, ON, Canada). We followed the method reported by Tanaka and Ludwig (47). Briefly, each strip was stretched at a rate of 0.2 mm/s until the sample separated into two pieces. Due to the limitation of the maximum displacement of the lever arm, the strips were first stretched to approximately three times their unstretched length before the displacement was recorded and force signals were started. The strain was defined as the total displacement in length normalized by the unstretched length of the samples. A transient decrease in the force (see arrows in Fig. 8) was defined as "the failure stress," indicating that fibers in the alveolar walls started to break during stretching. Two or three strips were prepared and measured from each animal, and the average value of the failure stress was used for statistical analysis.

**Bronchoalveolar lavage.** Whole lung lavages were performed with 1 ml of PBS twice via the tracheal cannula after impedance and P-V curve measurements were completed (n = 5 in each group). The return volume was measured, and the bronchoalveolar lavage fluid (BALF) was centrifuged. The cell pellet was resuspended in red blood cell lysis buffer (0.01% NH₄Cl) and brought up to the initial lavaged return volume for total cell count by hemocytometry. Slides contain-
ing 500–1,000 cells were prepared using a cytocentrifuge and stained with rapid Wright’s stain. Differential cell analysis was performed by manual counting under a light microscope.

**Lung histology and morphometry.** After mediastinal dissection, the lungs were perfused with 10% buffered formalin via the tracheal cannula at an airway pressure of 25 cmH₂O for at least 20 min (n = 5 in each group). The fixed lungs were embedded in paraffin, sectioned, and stained with hematoxylin and eosin for histological analyses. The average distance between alveolar walls, the mean linear intercept (MLI), was calculated according to established methods (10) using a light microscope. For each pair of lungs, 10 histological fields were evaluated.

**BALF elastase-like activity.** Elastase-like activity in BALF was measured by following the protocol of Dhami et al. (9). Briefly, BALF samples were lyophilized and reconstituted in water to make a fivefold concentrated solution. An assay buffer of 0.2 M Tris·HCl, pH 8.0, was prepared; 100 μl of assay buffer, 50 μl of substrate (0.5 mg/ml of N-succinyl-Ala-Ala-Ala-p-nitroanilide), and 50 μl of BALF were then added. All samples were assayed in duplicate. EDTA (10 mM) was added to the samples to inhibit metalloelastase (13). Negative controls of 150 μl of assay buffer and 50 μl of substrate were used. We assessed background absorbance of each BALF sample by incubating 150 μl of assay buffer with 50 μl of each sample. This value was then subtracted from the absorbance of the test wells. We measured the absorbance of the wells at 405-nm wavelength using a spectrophotometer.

**Whole lung collagen content.** Collagen content was assessed by measuring the hydroxyproline content of the tissue as previously described by Woessner (51). Lungs from the control (n = 7) and the PPE-treated (n = 7) mice were lyophilized for 12 h to dry weight, measured, and minced. The lung sample was then hydrolyzed with 4 ml of 6 N HCl at 100°C for 6 h. One milliliter of the hydrolysate was then taken and evaporated. The powder was reconstituted with 1 ml of distilled H₂O and reevaporated. The powder was then reconstituted with 5 ml of distilled H₂O. Hydroxyproline (Sigma) standard solutions of 0–10 μg/ml were prepared. Sample solution (2 ml) was taken and oxidized with 1 ml of chloramine-T (Sigma) for 20 min. The reaction was then stopped with 1 ml of 3.15 M perchloric acid. After 5 min, 1 ml of p-dimethylaminobenzaldehyde solution was added. The sample was vortexed, incubated in a 60°C bath, and then cooled under tap water for 5 min. The absorbency of the solutions was determined at 557 nm using a spectrophotometer. The hydroxyproline concentration was determined from the standard curve.

**Whole lung elastin levels.** The lyophilized samples were placed into 1 ml of 0.25 M oxalic acid. The suspension was then heated at 100°C for 1 h. The specimen was centrifuged, and the supernatant was collected. The above procedure was repeated for a total of five times until all the insoluble elastin had been converted into a soluble product (α-elastin) (7). One milliliter of the collected supernatants for each mouse was then dialyzed against water using 15,000 molecular-weight cutoff dialysis membrane (Spectrum, Houston, TX). The levels of α-elastin were then determined with the use of Fastin-elastin assay (Biocolor, Belfast, Northern Ireland) following the specific protocol outlined in the kit.

**Statistical analysis.** All data were expressed as means ± SD. Student’s t-test and repeated-measures two-way ANOVA were used to evaluate the significance of differences between means and variances, with P < 0.05 as the level of significance.

**RESULTS**

**Histopathology.** Figure 1 shows representative alveolar structures of a control and a PPE-treated lung stained with hematoxylin and eosin 3 wk after treatment. Significant enlargement of the alveolar air spaces was observed in lung samples from PPE-treated mice compared with control mice. Both the mean and the SD of the MLI in the PPE-treated group (83 ± 21 μm) were significantly larger than in the control group (43 ± 3 μm) (P = 0.001). There were no significant differences in body weight between PPE-treated (24.9 ± 0.9 g) and control groups (25.2 ± 1.1 g).

**BALF analysis.** There were no significant differences in BALF total cell numbers between PPE-treated (3.4 ± 1.0 × 10⁴ /ml) and control groups (3.8 ± 0.9 × 10⁴ /ml), and over 95% of the cells were macrophages in both groups. No eosinophils and only 0.5% neutrophils were observed in both groups. Additionally, no elastase-like activity was detected in BALF of the control and PPE-treated mice, suggesting that direct elastolytic activity of PPE had already diminished at 3 wk after treatment as reported previously (40).

**P-V curves and dynamic respiratory mechanics.** Figure 2 compares the quasi-static P-V curves in the two groups obtained during volume-controlled inflation from end-expiratory volume. The pressure was significantly lower in the PPE-treated mice than in the control mice (P < 0.001). The average
quasi-static elastance value, defined as a slope between 0 and 1.15 ml of inflated volume, was significantly lower in the PPE-treated mice (18.8 ± 2.8 cmH2O/ml) than in the control mice (26.5 ± 1.7 cmH2O/ml) (P < 0.001).

Figure 3 shows representative cases of the dynamic respiratory system resistance and elastance (calculated from the reactance where elastance = −2πf × reactance) as a function of frequency and the fits of the mathematical model (20) to the data in representative control and PPE-treated mice at PEEP = 3 cmH2O. The measured data were fit well by the model at all PEEP levels in both groups. As a function of PEEP, all the values of tissue H parameters, H_min, H_max and H_mean, were significantly PEEP dependent (P < 0.001) and decreased with increasing PEEP in both groups (Fig. 4). The values of H_min, H_max and H_mean were significantly lower in the PPE-treated group than in the control group (P < 0.001) (Fig. 4). The largest difference in H_max between the control and the PPE-treated mice was at the highest (9 cmH2O) PEEP. Hysteresivity (η) was significantly PEEP dependent in the PPE-treated mice (P < 0.001) but not in the control mice and was significantly higher in the PPE-treated mice at PEEP ≥ 6 cmH2O (Fig. 5A). Raw was significantly PEEP dependent (P < 0.001), but there was no significant difference between the groups (P = 0.78) (Fig. 5B). G was significantly PEEP dependent (P < 0.001) and significantly lower in the PPE-treated mice (P < 0.001) (Fig. 5C).

Dynamic nonlinearity. The k_d, a measure of dynamic nonlinearity of the respiratory system, was significantly PEEP dependent (P < 0.001) and decreased, implying more linear behavior when PEEP was increased in both groups (Fig. 6). The k_d was significantly larger in the PPE-treated group than in

Fig. 2. Mean ± SD of quasi-static inspiratory pressure-volume relationships from end-expiratory lung volume of the control and the PPE-treated mice. Values are means ± SD. *P < 0.05.

Fig. 3. Representative examples of respiratory system resistance (A) and elastance (B) of the control and the PPE-treated mice calculated from the impedance data and corresponding fits of the model to the control and PPE-treated data at a positive end-expiratory pressure (PEEP) of 3 cmH2O.

Fig. 4. Dynamic tissue elastance (H) parameters as a function of PEEP. Means ± SD of minimum H (H_min) and maximum H (H_max) (A) and mean H (H_mean; B) in the control and PPE-treated mice. *P < 0.05.
the control group ($P < 0.001$). Figure 7 shows the linear relationships between $H_{\text{min}}$ or $H_{\text{max}}$ and $k_d$, including data from all PEEP levels. There were significant correlations between $H_{\text{min}}$ and $k_d$ in both the PPE-treated ($k_d = 0.51 \frac{H_{\text{min}}}{H_{\text{max}}} - 1.56$) ($P < 0.001$, $r = 0.83$) and control groups ($k_d = 0.36 \frac{H_{\text{min}}}{H_{\text{max}}} - 2.21$) ($P < 0.001$, $r = 0.90$) (Fig. 7A). The slopes of the two lines were significantly different ($P < 0.05$). There were also significant correlations between $H_{\text{mean}}$ and $k_d$ control mice ($k_d = 0.14 \frac{H_{\text{max}}}{H_{\text{max}}} - 0.82$) ($P < 0.001$, $r = 0.86$) (Fig. 7B). The two lines were nearly parallel, and there was no significant difference in slopes between the two lines ($P = 0.48$); however, the intercept of the PPE-treated group was relatively higher, nearly reaching a statistical level ($P = 0.07$). There were also significant correlations between $H_{\text{mean}}$ and $k_d$. 

Fig. 6. Harmonic distortion ($k_d$), an index of dynamic nonlinearity of the respiratory system, of the control and the PPE-treated mice as a function of PEEP. *$P < 0.05$.

Fig. 7. Correlations between $H_{\text{min}}$ (A) or $H_{\text{max}}$ (B) and $k_d$. Data were obtained from of the control ($n = 8$) and the PPE-treated ($n = 9$) mice at PEEP = 0, 3, 6, and 9 cmH2O.

Fig. 5. Means ± SD of hysteresivity (A), airway resistance (Raw; B), and tissue damping (G; C) in the control and PPE-treated mice as a function of PEEP. *$P < 0.05$. 

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in both the PPE-treated \( (k_d = 0.29 H_{\text{mean}} - 2.04) \) \( (P < 0.001, r = 0.93) \) and the control groups \( (k_d = 0.23 H_{\text{mean}} - 3.00) \) \( (P < 0.001, r = 0.94) \), and the slopes of the two lines were significantly different \( (P < 0.05) \) (data not shown). After the PPE treatment, each point moved to the left at an equivalent PEEP, implying stronger nonlinearity per unit tissue elastance in the PPE-treated group.

**Lung tissue failure tests.** Figure 8 shows representative traces of parenchymal tissue strip failure tests from a PPE-treated and a control mouse. The stress in each tissue strip increased with strain until a maximum was reached. Next, the stress started to drop in discrete steps, suggesting that groups of alveolar walls broke within the tissue. Finally, the stress levels returned to zero when the samples separated into two pieces.

Table 1 shows the summary of failure test results for all parenchymal tissue strips. The stress that developed just before the strips started failing, defined as the failure stress, was significantly lower by 40% in the PPE-treated group than in the control group \( (P = 0.002) \), demonstrating that the PPE-treated parenchymal tissues were weaker than control tissues. There were no significant differences in the strain values at failure \( (P = 0.56) \).

**Lung hydroxyproline and elastin levels.** The levels of hydroxyproline and \( \alpha \)-elastin are expressed as microgram per milligram dry weight of lung tissue (Table 2). There were no significant differences in dry lung weight between the control and the PPE-treated mice \( (P = 0.81) \). The levels of hydroxyproline, a measure of total collagen, were significantly higher in the PPE-treated mice, by 48%, than in the control mice \( (P = 0.002) \). The levels of \( \alpha \)-elastin were lower by 13% in PPE-treated mice, but the difference was not statistically significant \( (P = 0.16) \).

**DISCUSSION**

The primary findings of this study are that after PPE treatment of mice 1) lung elastance decreased and hysteresivity increased, 2) parenchymal tissue fibers failed at a lower stress than shown in the control group, 3) dynamic nonlinearity characterized by \( k_d \) increased, and 4) total collagen content increased, whereas there was a tendency of loss of elastin of whole lungs in the emphysema group. These physiological observations support the hypothesis that exposure of the lung to PPE leads to a remodeling of the fiber network that significantly alters the mechanical properties and the nonlinear mechanical behavior of the whole organ.

**Collagen remodeling in emphysema.** In emphysema, the loss of elastin from the alveolar walls appears to be a major event in clinical pathology (1). The breakdown and the degradation of collagen fibers have also been reported in both human patients (39) and rodent models of emphysema (9, 35, 52). In addition, several studies have demonstrated that, after the destruction of alveolar walls, remodeling of collagen fibers as a result of an abnormal repair process contributes to the pathogenesis of emphysema (12, 17, 21, 22, 25, 49). Increases in total amount of collagen of the lungs have been reported in human patients (23, 35). In the present mouse model of emphysema, we also observed a 45% statistically significant increase in total collagen content and a 13% decrease in total elastin content of the whole lung (Table 2), suggesting that collagen remodeling within the lung was indirectly triggered after the elastolytic injury. Previous studies have shown that, after the onset and initial progression of emphysema due to the proteolytic injury caused by PPE, synthesis of collagen by lung fibroblasts is considered to be upregulated as part of the repair process of the damaged lung (15, 22). However, the repair process does not restore normal structure and function to the lung leading to pathology and altered physiology (24). Because the extracellular assembly of collagen molecules to fibrils and fibers is sensitive to the composition of the surrounding matrix (22, 24), it is conceivable that the structure and mechanical properties of the newly synthesized collagen differ from those that occur during normal growth. Although the details of this abnormal repair process at the molecular level are beyond the scope of the present study, it is important to discuss the physiological consequences.

**Alveolar wall and fiber failure in emphysema.** It has long been proposed that mechanical failure of the alveolar walls plays a pivotal role in the progression of emphysema (50). Recently, Kononov et al. (21) observed the failure of a single alveolar wall in a rat model of PPE-induced emphysema. However, to our knowledge, this is the first study to quantify the failure stress of parenchymal tissue strips from emphysematous lungs. We found evidence that the emphysematous

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**Table 1. Tissue fiber rupture test**

<table>
<thead>
<tr>
<th></th>
<th>Failure Stress, kPa</th>
<th>Strain</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control</td>
<td>11.9±1.4</td>
<td>3.33±0.49</td>
</tr>
<tr>
<td>PPE-treated</td>
<td>7.2±2.4</td>
<td>3.59±0.93</td>
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</tbody>
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\( P = 0.002 \) \( P = 0.56 \)

**Values are means ± SD \( (n = 6 \) mice/group).** PPE, porcine pancreatic elastase.

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**Table 2. Lung collagen and elastin contents**

<table>
<thead>
<tr>
<th>Dried Lung Weight, mg</th>
<th>Hydroxyproline, ( \mu )g/mg</th>
<th>( \alpha )-Elastin, ( \mu )g/mg</th>
</tr>
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<tbody>
<tr>
<td>Control</td>
<td>13.5±3.8</td>
<td>4.38±0.83</td>
</tr>
<tr>
<td>PPE-treated</td>
<td>13.1±2.4</td>
<td>6.38±1.33</td>
</tr>
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**Values are means ± SD.** Hydroxyproline and \( \alpha \)-elastin contents are expressed as \( \mu \)g/mg of dried lung weight \( (n = 7 \) mice/group).
lung tissue breaks at the same strain but at a stress 40% smaller than the normal tissue (Table 1). In an attempt to interpret these results, we first note that normal collagen fibers are stiffer and stronger than other connective tissue constituents (36). As a consequence, the amount and organization of collagen in the alveolar walls should play a crucial role in determining the stiffness and the failure properties of the lung tissue. For example, when the collagen content of normal lung tissue strips was decreased via in vitro digestion using collagenase, the stiffness of the tissue dropped to 40% of its value before digestion (53). Alternatively, normal developmental changes of the lung increase both the stiffness and the failure strength of the alveolar walls as well as collagen contents during maturation. Indeed, Tanaka and Ludwig (47) reported that the failure stress of normal lung tissue from baby rats was 18 kPa, which increased to 28 kPa in adult rats. At the same time, in a followup study, Tanaka et al. (46) also found that the collagen content of the lung increased from 18% to ~38% during this normal maturation process. Thus, because in this model of emphysema we found a 45% increase in lung collagen (Table 2), one would expect the stiffness and the failure stress of the tissue to increase. Surprisingly, however, our data showed just the opposite behavior: despite the increase in collagen content, both lung elastance and the failure stress decreased by ~30% (Figs. 2 and 4) and 40% (Table 1), respectively, indicating that the total amount of collagen in the lung tissue is not the primary determinant of the mechanical properties in the diseased state.

In an attempt to resolve the apparent contradiction between increased collagen content and decreased elastance and failure stress in the emphysematous tissue, we first discuss the relation of lung stiffness and alveolar structure. Because the MLI nearly doubled in the emphysematous group, the number of alveolar walls per unit volume that can resist the deformation of the tissue strip could have decreased. This mechanism alone could account for the lower elastance and lower failure stress even if the mechanical properties of the alveolar walls were similar to those of the normal tissue. However, it is likely that not all of the increase in MLI is due to alveolar wall rupture. The PPE-treated lung is softer, and, at the fixation pressure of 25 cmH2O, the alveoli would be more extended than those in the normal lung. Thus the number of alveolar walls per unit volume in the tissue strip is not necessarily smaller in the PPE-treated lung than in the normal lung. Using microscopic imaging, Brewer et al. (3) recently reported that individual alveolar walls from PPE-treated rats, which also involved collagen remodeling (21), appeared softer and more extensible than those from normal rats. These observations suggest that, despite the increased collagen content, the alveolar walls and the collagen fibers are likely to be weaker in the emphysematous lung as a consequence of the process of degrading and remodeling. Indeed, the ultrastructure of collagen from human emphysematous lungs reveals thickened and disorganized fibrils after remodeling (12). Our data then suggest that the stiffness and the failure properties of the remodeled fibers must decrease compared with normal collagen fibers. The reduction in the failure stress of collagen has an important effect on how the structure of the lung evolves during the progression of emphysema. Suki et al. (44) recently developed a fiber network model and argued that, because mechanical forces influence the process of tissue breakdown, the alveolar structure must be very heterogeneous and the alveolar walls around the perimeter of severe emphysema lesions or the walls that separate such lesions may be overstretched. In agreement with these predictions, the heterogeneity of the alveolar dimensions was found to be much larger in the emphysematous than in the control lungs, both in the present study and in previous studies (10, 20, 34). Therefore, the alveolar walls in the emphysematous lung may have to oppose larger stresses locally, and, as a result, the increased local stresses can promote rupture of the remodeled walls (44), which in turn results in a decreased failure stress, as observed in Table 1. We thus conclude that mechanical forces are expected to play an important role in the progression of emphysema once the collagen matrix has undergone a critical amount of remodeling.

Lung mechanical properties. The consequences of alterations in the ECM of alveolar walls can be traced to organ-level changes in the mechanical properties of the lung. Because the chest wall is very soft in the mouse, at least 90% of H is due to the lung parenchyma (37). Additionally, any change in H must be related to a change in lung mechanics; hence, in the discussion that follows, we assume that changes in H largely reflect changes in lung mechanics. We have recently developed a new mathematical model that assumes a continuous distribution of H between a minimum and a maximum value (Hmin and Hmax, respectively) (20). We found that all H-related parameters (Fig. 4) as well as the static elastance (Fig. 2) decreased in the emphysematous mice compared with controls. The Hmax represents the stiffest regional elastance in the lung, and the collagen should be the most important determinant of its value. Thus the lower Hmax values in the PPE-treated mice suggest that the ultrastructural changes of remodeled collagen weaken the fibers and the alveolar walls, in agreement with analyses of the failure tests. On the other hand, we speculate that Hmin represents the softest regional elastance, which may be related to the loss of alveolar walls in that region. Thus it is likely that the lower value of Hmin is a functional consequence of the increased MLI in the emphysematous mice.

The hysteresivity is a material property of the lung tissue (14), and it also depends on the microscopic constituents of the alveolar walls. Indeed, changes in ECM composition can cause a change in the hysteresivity in the parenchymal tissue level (33, 53). In the present study, hysteresivity of the PPE-treated mice was higher than that of the control mice, as observed in TGF-α transgenic emphysematous mice (32), in mild emphysematous mice induced by nebulized PPE-treatment (20), and in rats (3). In parenchymal tissues of normal guinea pigs, hysteresivity after in vitro digestion with collagenase was significantly higher than that after digestion with elastase (53). This suggests that, in the normal lung tissue, the larger the elastin-to-collagen ratio the larger the value of hysteresivity. However, compared with controls, hysteresivity of the emphysematous mice increased (Fig. 5A), whereas collagen content also increased (Table 2). Together, the hysteresivity and elastance results suggest that remodeling in emphysema produces weak and viscous alveolar walls that also fail at lower stresses than those of the normal lung.

The values of Raw decreased with increasing PEEP most likely due to the increasing diameters of the airways with lung inflation in both groups (Fig. 5B). Although increased Raw values, which suggest an underlying airway obstruction, were reported in sheep with experimental emphysema after papain...
treatment (19). Raw values were the same between the groups, perhaps because the effects of PPE treatment on the airways used in this murine model is distinct from that of papain that was used in the ovine model. To our knowledge, increases in Raw have not yet been reported in mouse models of emphysema. In another emphysema model of surfactant protein D-deficient mice, Raw values became lower than those shown in controls (5). Thus the physiological feature of the present model is similar to the classical physiological alterations in patients with α1-antitrypsin deficiency (2), in which airway conductance was within normal limits and the primary physiological defect was a loss of elastic recoil.

Because surfactant plays an important role in the mechanical properties of the normal lung, it is conceivable that alterations in surfactant properties in the emphysematous lung could influence our results. Because fewer type II pneumocytes that secrete surfactant were observed in the lungs of human emphysema patients (31), it has been suggested that surfactant plays a protective role against the development of pulmonary emphysema. In PPE-treated mice, administration of surfactant prevents the development of emphysema (30). However, it is not known whether the composition and biophysical properties of lung surfactant change as a consequence of the development of emphysema. Although the expression of surfactant protein A messenger RNA is increased in the lungs of the klotho mouse model of emphysema (41), this protein plays a less important role in stabilizing the alveoli than the hydrophobic surfactant proteins (18). One could argue that, due to the reduction in lung tissue recoil, surface tension may become even more important in emphysema than in the normal lung. However, emphysema is associated with pronounced heterogeneity at the alveolar level, and it is unclear whether abnormalities in surfactant actually contribute to the development of the disease. The extent to which lung surfactant contributes to recoil in emphysema has also not been well characterized. Further studies would be needed to clarify the role of surfactant in emphysema.

Dynamic nonlinearities. Another important physiological finding of this study is that the tissues responsible for generating elastic recoil in the emphysematous lungs also displayed significantly greater nonlinear behavior than control lungs. The mechanical behavior of the normal lung tissue has been characterized as nonlinear (28, 29, 42, 53), and the origin of dynamic nonlinearity has been investigated in various organs (11). In the respiratory system, dynamic nonlinearity is likely related to the ECM components, including the nonlinearly viscoelastic collagen and its interactions with the linear viscoelastic elastin, and the viscous ground substance, including mainly proteoglycans (29). Because elastic fibers behave more linearly than collagen fibers (28, 36), tissue nonlinearity could be more related to collagen and, in particular, the extent to which collagen fibers are stretched in the alveolar walls. Thus the dynamic nonlinear behavior of the lung tissue can be considered as a global in vivo assay of collagen function in the intact lung. Although the nonlinearity is certainly related to collagen, it is also possible that the physical interaction between collagen and elastin also influences nonlinear behavior.

In the present study, we demonstrated for the first time that dynamic nonlinearity of the whole lung, as characterized by $k_d$ (42), is linearly related to lung elastance parameters ($H_{\text{min}}$ and $H_{\text{max}}$) in both the PPE-treated and the control mice (Fig. 7), as found for normal tissue strips (53). More importantly, despite a decrease in $H$, $k_d$ increased with emphysema compared with normal lungs. Specifically, the relationship between $k_d$ and $H_{\text{min}}$ as well as $H_{\text{max}}$ shifted to the left in emphysema. This can be accounted for by changes in ECM components. As discussed above, a decrease in elastance and an increase in collagen content suggest that the new collagen in the remodeled alveolar walls must be less stiff than the normal collagen. An increase in $k_d$ on the other hand suggests that the collagen fibers are either more stretched or inherently different from normal with respect to their nonlinear mechanical behavior in the emphysematous alveolar wall.

One may argue that, in contrast to tissue strips, in the whole lung, airflow closure also contributes to harmonic distortion. If a significant portion of the lung is blocked by airway closure and tidal volume remains the same, then a smaller lung will receive the same tidal volume and the lung becomes over-stretched. In fact, the $k_d$ was largest at PEEP = 0 in both groups (Fig. 6), which is the condition where recruitment and derecruitment could occur most during oscillations. Furthermore, the decrease in $k_d$ with PEEP implies gradual recruitment. However, neither lung elastance nor $k_d$ decreased when PEEP was increased from 6 to 9 cmH2O. Thus these data suggest that, above 6 cmH2O PEEP, recruitment did not occur and hence it could not influence our data. Therefore, we believe that the relation between $k_d$ and $H$, and consequently the above interpretation of the results, is insensitive to airflow closure at least at the higher PEEPs included in this study.

In summary, we have characterized the respiratory and lung mechanical properties of a mouse model of emphysema induced by PPE. We observed a decrease in lung elastance and failure strength of the alveolar walls as well as an increase in hysteresivity, dynamic nonlinearity, and total lung collagen content. These results suggest that significant collagen remodeling takes place within the alveolar wall, which produces weak but more nonlinear fibers and alveoli that are locally over-stretched and hence prone to mechanical failure. These alterations in the micromechanics of the alveolar walls significantly affect organ-level lung function in the mouse.

**GRANTS**

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**REFERENCES**


