Effect of gravity on chest wall mechanics

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1Dipartimento di Medicina Sperimentale, Ambientale e Biotecnologie Mediche, Università di Milano-Bicocca, I-20052 Monza, Italy; 2Médecine Aerospatiale, Université de Bordeaux, F-33076 Bordeaux; 3Centre Chirurgical Marie Lannelongue, Unité Propre de Recherche de l’Enseignement Supérieur Equipe Associee 2397, Université Paris XI, F-92350 Le Plessis Robinson; and 3Hôpital Lariboisière, F-75010 Paris, France
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Bettinelli, D., C. Kays, O. Bailliart, A. Capderou, P. Techoueyres, J. L. Lachaud, P. Vaida, and G. Miserocchi. Effect of gravity on chest wall mechanics. J Appl Physiol 92: 709–716, 2002; 10.1152/japplphysiol.00644.2001.—Chest wall mechanics was studied in four subjects on changing gravity in the craniocaudal direction (Gz) during parabolic flights. The thorax appears very compliant at 0 Gz: its recoil changes only from −2 to 2 cmH2O in the volume range of 30–70% vital capacity (VC). Increasing Gz from 0 to 1 and 1.8 Gz progressively shifted the volume-pressure curve of the chest wall to the left and also caused a fivefold exponential decrease in compliance. For lung volume <30% VC, gravity has an inspiratory effect, but this effect is much larger going from 0 to 1 Gz than from 1 to 1.8 Gz. For a volume from 30 to 70% VC, the effect is inspiratory going from 0 to 1 Gz but expiratory from 1 to 1.8 Gz. For a volume greater than ~70% VC, gravity always has an expiratory effect. The data suggest that the chest wall does not behave as a linear system when exposed to changing gravity, as the effect depends on both chest wall volume and magnitude of Gz.

METHODS

The parabolic flights. All experiments were conducted during three European Space Agency-Centre National d’Etudes Spatiales campaigns of parabolic flights between October 1999 and April 2000. Each campaign included three flight days. Each flight was performed by an Airbus A300 and lasted 2.5–3 h, including 30 parabolas (on the whole, 90 parabolas per campaign). The parabolic flight allows change of the Gz vector relative to steady horizontal flight (1-Gz phase). During pull-up, an acceleration of 1.8 Gz is reached and maintained for ~20 s. Subsequently, reducing engine thrust allows the aircraft to enter a free-falling parabolic trajectory that generates a 0-Gz phase lasting ~20 s. Finally, during pull-out, another 1.8-Gz phase is reached lasting for ~20 s before the return to steady horizontal flight at 1 Gz.

Subjects. Respiratory mechanics were studied in four subjects, three men and one woman, during steady horizontal flight and during short periods of 0 Gz and 1.8 Gz. The same subjects were also studied in ground experiments in sitting and supine postures using the same equipment. The subjects were members of the experimental team. They were non-smokers, in good health, and had no report of pulmonary disease; their anthropometric features are given in Table 1. The subjects were trained to perform the respiratory maneuvers and, in particular, the “relaxation” maneuver, which is necessary to describe the volume-pressure features of the respiratory system. Furthermore, all subjects took part in

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previous parabolic flight campaigns and were well accustomed to the challenge of abruptly changing $G_z$ several times during each flight.

Experimenal equipment and system. Subjects were sitting in a body plethysmograph made of wood (empty volume of 360 liters) that was equipped with a pneumotachograph and transducers to measure pressure in the box and at the mouthpiece (Pm); the mouthpiece was also provided with an electromagnetic shutter. We also performed some parabolas with subjects off the transducers to evaluate the dependence of transducer signals from acceleration. To minimize the effect of changes in aircraft accelerations on both transducers, they were oriented along the aircraft’s transverse axis. Lung volumes were measured by flow integration; thoracic gas volume (TGV) was measured at the end of expiration by closing the mouthpiece shutter and performing inspiratory and expiratory efforts against closed airways (panting maneuver) for 3 s. Esophageal (Pes) and gastric pressures were measured by two standard pressure transducers mounted on a single GaëlTec CTO-2 catheter, with a 2-mm external diameter (9). Transducer sensitivity was $5 \mu V^{-1} \cdot mmHg^{-1}$; linear pressure range was $\pm 300 \text{ mmHg}$. Subjects advanced the catheter through the nose until the proper positioning of the esophageal probe was reached based on preliminary experiments aimed at determining the best recording site. This was decided in consideration of the minimal cardiac artifact and a stable pressure signal.

The pneumotachograph response was linear for flow rates compatible with the respiratory maneuvers performed; the maximum error was $\sim 5\%$ at high flow rates ($\sim 3 \text{l/s}$).

All signals were acquired through a system made of an analog-to-digital card (50 Hz/channel; Digimétrie), stored into a personal computer DOS “homemade” program that allowed “on-line” plethysmograph pressure conversion into pneumotachograph flow and volume by integration, and displayed, on a video screen, current lung volume, pressure variables, and $G_z$. The software also allowed us to process an “off-line” preliminary data analysis.

All of the data were also stored on an analog tape recorder as a backup, as were the live comments from the researchers during the flight.

Calibration. Before take off, calibration of the plethysmograph was done using a 2-liter syringe. A syringe volume control was made for each subject during the flight. Pressure transducer calibration for body box pressure and Pm was carried out using a water manometer.

GaëlTec transducers were calibrated by using a special calibration chamber in which pressure could be set by water manometers; sensitivity of transducers and zero drift at atmospheric pressures were carefully noted. Zero drift was slightly dependent on temperature. To account for body temperature with an in situ catheter, we measured the zero values on withdrawal of the probe at the end of each experimental session. These zero values were then used to correct both Pes and gastric pressures that were previously recorded.

Protocols for in-flight experiments. Subjects were sitting inside the plethysmograph and breathing through the mouthpiece. During 0-Gz exposure, there was a tendency for the subject to float up in the air because of the changing trajectory of the aircraft. To counteract such inertia-dependent phenomenon, the subject was kept strapped at the thighs and feet; other loose bands around the arms kept the arms in a natural position parallel to the chest. During level flight, a check was performed to ensure regular recording of all variables. The time frame for data acquisition during respiratory maneuvers started in the last minute of level flight and lasted 2 min as follows: level flight (1 $G_z$, 1 min), pull up (1.8 $G_z$, 20 s), injection (0 $G_z$, 20 s), and pull out (1.8 $G_z$, 20 s).

Subjects were instructed to perform timely, within each phase, different respiratory maneuvers, as detailed below, that were always preceded by a few control breaths followed by TGV measurement around the end-expiratory volume. Occasionally, TGV was also measured at the end of the 2-min time frame.

VC maneuver. Subjects inspired up to total lung capacity (TLC) and expired slowly down to residual volume (RV).

The relaxation maneuver. The respiratory system is made of two components, the chest wall and the lung, placed mechanically in parallel (1). Accordingly, when the respiratory muscles are relaxed, the total pressure exerted by the respiratory system at alveolar level ($P_A$) is given by $P_A = P_w + P_t$, where $P_w$ is the pressure exerted by the chest wall, and $P_t$ is the pressure exerted by the lung. Recalling that $P_t = P_{w} - P_{pl}$, where $P_{pl}$ is pleural surface pressure estimated from Pes, and substituting into the equation above, one has, at a given lung volume when respiratory muscles are relaxed, $P_w = P_{pl}$. To measure Pes in the relaxed condition, the subject has to reach a given lung volume, either inspiring or expiring relative to the end-expiratory volume, and then relax the respiratory muscles, either against a closed mouthpiece shutter or by voluntarily closing the glottis. The maneuver requires some practice, and, in a well-trained subject, it takes a few seconds to reach a stable Pes value.

Only one VC maneuver or one relaxation maneuver could be performed during each $G_z$ phase. To evaluate whether relaxation was adequate, subjects were also asked to relax toward the end of a given $G_z$ phase and to keep the relaxation during the early part of the following $G_z$ phase. In fact, considering that the change in $G_z$ occurs in $2–3$ s, we hypothesized that a change in Pes paralleling $G_z$, with a final setting to a new steady Pes value, would be in favor of a good relaxation.

Total numbers of parabolas necessary to gather a complete set of data ranged, for each subject, from 15 to 20.

Protocols for ground experiments. Ground experiments were performed on the same subjects, adopting the same general protocols and equipment in sitting and supine postures. The change in posture was obtained by leaning the plethysmograph backward. This implied that legs remained inside the plethysmograph and breathing through the mouthpiece. During 0-Gz exposure, there was a tendency for the subject to float up in the air because of the changing trajectory of the aircraft. To counteract such inertia-dependent phenomenon, the subject was kept strapped at the thighs and feet; other loose bands around the arms kept the arms in a natural position parallel to the chest. During level flight, a check was performed to ensure regular recording of all variables. The time frame for data acquisition during respiratory maneuvers started in the last minute of level flight and lasted 2 min as follows: level flight (1 $G_z$, 1 min), pull up (1.8 $G_z$, 20 s), injection (0 $G_z$, 20 s), and pull out (1.8 $G_z$, 20 s).

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Protocols for ground experiments. Ground experiments were performed on the same subjects, adopting the same general protocols and equipment in sitting and supine postures. The change in posture was obtained by leaning the plethysmograph backward. This implied that legs remained as in the sitting posture. To estimate a possible influence of the position of the legs, we also measured RV, VC, and TLC, maintaining the plethysmograph at an angle of $30^\circ$ relative to horizontal.

Data analysis. TGV, computed from Boyle’s law, is given by the following: $TGV = P_{w} \cdot (\Delta V/\Delta Pm)$, where $P_{w}$ is in-flight cabin pressure and $\Delta V/\Delta Pm$ is the ratio of change in TGV to the change in Pa during panting maneuvers. This ratio was inferred as the slope of the linear regression between the two variables. Before the regression was executed, the drift affecting the volume of the panting maneuvers was subtracted;

### Table 1. Anthropometric features of the subjects studied

<table>
<thead>
<tr>
<th>Subject</th>
<th>Gender</th>
<th>Age, yr</th>
<th>Weight, kg</th>
<th>Height, cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>Male</td>
<td>55</td>
<td>82</td>
<td>173</td>
</tr>
<tr>
<td>B</td>
<td>Female</td>
<td>47</td>
<td>47</td>
<td>161</td>
</tr>
<tr>
<td>C</td>
<td>Male</td>
<td>58</td>
<td>75</td>
<td>173</td>
</tr>
<tr>
<td>D</td>
<td>Male</td>
<td>51</td>
<td>72</td>
<td>176</td>
</tr>
</tbody>
</table>
therefore, we generally obtained regression coefficients near 0.99, thus ensuring an accurate TGV measurement.

Lung volume recorded throughout the time frame also displayed a drift because of increasing temperature inside the plethysmograph. Digital reading of lung volumes was obtained after the volume drift between two successive TGV values was corrected, assuming a linear drift with time. Pes data were corrected for the zero drift on withdrawal of the catheter at the end of the session. A “moving-average filter” was employed to reduce high-frequency noise in the pressure records.

Pressure data obtained during relaxation maneuvers were considered acceptable when they remained steady during the maneuver. At 1.8 Gz, whenever possible, we preferred data gathered during the pull-up phase, as the Gz vector remained more steady.

For each subject, we plotted the lung volume, expressed as %VC, against relaxation Pw. With the use of a curve-fitting procedure (see APPENDIX), the individual volume-pressure relationships at 1, 1.8, and 0 Gz, and in the supine position were determined. We defined, as resting volume of the thorax, the volume at which its recoil pressure is zero, namely the intercept of the volume-pressure relationship on the y-axis.

Finally, a statistical analysis was carried out to verify statistical significance of results regarding VC, TLC, RV, compliance, and resting volume of the chest wall. Two types of tests were executed: a paired t-test and an ANOVA test for repeated measures followed by the Student-Newman-Keuls posttest (95% confidence interval).

RESULTS

Examples of relaxation maneuvers at volumes above the end-expiratory volume are presented in Fig. 1 at 1, 1.8, and 0 Gz. In the records of the Pm and Pes, one can easily detect the oscillations referring to the panting maneuver performed to measure the total gas volume. After the panting maneuver was completed, the subject inspired, closed the glottis, and relaxed. The lung volume remained constant, and Pes reached a steady value that represented the recoil elastic Pw.

Figure 2A shows two examples of a relaxation maneuver maintained on changing Gz. In Fig. 2A, the relaxation was performed at a volume below the end-expiratory volume during the last phase of 1.8 Gz and extended to 0 Gz. Pes increased, sharply synchronized to the decrease in Gz, and remained steady thereafter. In Fig. 2B, an example of the relaxation maneuver...
Table 2. Residual volume, vital capacity, and total lung capacity of the subjects in the various experimental conditions

<table>
<thead>
<tr>
<th>Subject</th>
<th>In Flight</th>
<th>On Ground</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>0 Gz</td>
<td>1 Gz</td>
</tr>
<tr>
<td></td>
<td>1 Gz</td>
<td></td>
</tr>
<tr>
<td>RV, ml</td>
<td></td>
<td></td>
</tr>
<tr>
<td>A</td>
<td>2,100 ± 110 (2)</td>
<td>2,250 ± 30 (3)</td>
</tr>
<tr>
<td>B</td>
<td>1,860 ± 40 (3)</td>
<td>1,830 ± 40 (3)</td>
</tr>
<tr>
<td>C</td>
<td>3,080 ± 140 (4)</td>
<td>3,100 ± 190 (4)</td>
</tr>
<tr>
<td>D</td>
<td>1,750 (1)</td>
<td>1,790 ± 170 (2)</td>
</tr>
<tr>
<td>Mean ± SE</td>
<td>2,200 ± 300</td>
<td>2,240 ± 300</td>
</tr>
<tr>
<td>VC, ml</td>
<td></td>
<td></td>
</tr>
<tr>
<td>A</td>
<td>5,570 ± 70 (2)</td>
<td>5,460 ± 90 (3)</td>
</tr>
<tr>
<td>B</td>
<td>3,460 ± 30 (3)</td>
<td>3,400 ± 40 (3)</td>
</tr>
<tr>
<td>C</td>
<td>5,500 ± 270 (4)</td>
<td>6,080 ± 100 (4)</td>
</tr>
<tr>
<td>D</td>
<td>5,430 (1)</td>
<td>5,710 ± 20 (2)</td>
</tr>
<tr>
<td>Mean ± SE</td>
<td>4,990 ± 510</td>
<td>5,160 ± 600</td>
</tr>
<tr>
<td>TLC, ml</td>
<td></td>
<td></td>
</tr>
<tr>
<td>A</td>
<td>7,670 ± 180 (2)</td>
<td>7,710 ± 60 (3)</td>
</tr>
<tr>
<td>B</td>
<td>5,320 ± 40 (3)</td>
<td>5,230 ± 80 (3)</td>
</tr>
<tr>
<td>C</td>
<td>8,580 ± 370 (4)</td>
<td>9,180 ± 190 (4)</td>
</tr>
<tr>
<td>D</td>
<td>7,190 (1)</td>
<td>7,500 ± 150 (2)</td>
</tr>
<tr>
<td>Mean ± SE</td>
<td>7,190 ± 690</td>
<td>7,400 ± 810</td>
</tr>
</tbody>
</table>

Values are means ± SE. Nos. in parentheses, no. of measurements. RV, residual volume; VC, vital capacity; TLC, total lung capacity; Gz, gravity in the craniocaudal direction. *Significance relative to 1 Gz (on ground), one-way ANOVA for repeated measures, P < 0.05.

above the end-expiratory volume during a similar change in Gz is shown. In this case, a marked decrease in Pes paralleled the decrease in Gz.

Table 2 summarizes the data of RV, VC, and TLC. The only significant changes observed were a decrease in VC in the supine posture and in 30° head-up tilt, accounting for a similar decrease in TLC. No significant difference was found in RV, VC, and TLC between supine and 30° head-up tilt.

In Fig. 3, the individual volume-pressure curves of the thorax are presented for the four subjects studied at 1, 1.8, and 0 Gz and in the supine posture. In all of the subjects, the compliance of the chest wall in the volume range of 30–70% VC increased significantly going from 1.8 to 1 Gz to supine, and to 0 Gz (Table 3). Relative to 1 Gz, the resting volume of the thorax decreased significantly (Table 3) in the supine posture, at 0 Gz, and at 1.8 Gz (−33%, −20%, and −8.4% VC, respectively).

Figure 4 shows the average curves for all of the subjects. They were obtained by reading and averaging volumes at the isopressure value, thus considering

Fig. 3. Individual volume-pressure relationships of the 4 subjects studied at 1 (solid line), 1.8 (dotted line), and 0 Gz (dashed line) and in supine posture (dotted-dashed line). Volumes are expressed as %vital capacity (VC). Patm, atmospheric pressure.
pressure as the independent variable (the average compliance values are reported in Table 3).

Figure 5 presents the compliance values of the four subjects as a function of $G_z$. The dependence of compliance from $G_z$ could be well described by a simple exponential of the type $C = C_{lim} + (C_0 - C_{lim})e^{-G_z/G}$, where $C$ is compliance, $C_{lim}$ is the minimum achievable value of chest compliance, $C_0$ is the compliance at 0 $G_z$, and $G$ is a coefficient expressing the dependence of the chest wall compliance on $G_z$. The average value of compliance for the four subjects at 1.8 $G_z$ was 0.15 l/cmH$_2$O (Table 3), very close to the average value of $C_{lim}$, namely, 0.14 l/cmH$_2$O. The compliance values corresponding to the supine posture were plotted on each subject’s curve. This allows the estimation of the effect of supine posture on chest wall compliance to be, on the average, comparable to the effect of 0.45 $G_z$ in the sitting posture.

**DISCUSSION**

The opportunity to take part in parabolic flight campaigns offered the clue to evaluate the effect of changing the gravity vector in the craniocaudal direction ($G_z$ equal to 0, 1, and 1.8) on the mechanical properties of the chest wall. The subjects taking part in the study were well trained, “good relaxers” and, as also supported by the relaxation maneuvers during changing $G_z$, were able to attain adequate relaxation during the 20 s of 1.8-$G_z$ and 0-$G_z$ exposure. Therefore, the relaxation data should be considered satisfactory, despite the fact that electromyographic activity of the scalene and parasternal muscles may be present on changing $G_z$ (6, 14). We, therefore, used the relaxation data to construct volume-pressure curves of the chest wall to derive the dependence of chest wall compliance on gravity.

**What is the effect of gravity on the chest wall?** The relative volume contribution of the rib cage and of the diaphragm (abdominal contribution) at lung volumes around end-expiratory volume has been previously studied by inductive plethysmography (3, 16). When subjects went from 0 to 1 $G_z$, a decrease in rib cage volume was observed (expiratory action), concomitant to an increase in abdominal contribution because of caudal displacement of the diaphragm (inspiratory action). Because the latter was larger than the former, the overall resultant effect was found to be inspiratory.

When subjects went from 1 to 1.8 $G_z$, at a volume corresponding to ~40% VC, inductive plethysmography showed a modest inspiratory effect, mainly be-

![Fig. 4. Average volume-pressure relationships of the 4 subjects studied at 1 (solid line), 1.8 (dotted line), and 0 $G_z$ (dashed line) and in supine posture (dotted-dashed line). Volumes are expressed as %VC. Values are means ± SE.](image)

![Fig. 5. Chest wall compliance (C) vs. $G_z$ relationships. The points referring to the supine posture (open symbols) were plotted on the relationships of each subject to estimate the $G_z$ factor (solid symbols) equivalent to the gravity acting in the anteroposterior direction in the supine posture. Inset: coefficients of the exponential fitting equation, where $C_{lim}$ is the minimum achievable value of chest compliance, $C_0$ is the compliance at 0 $G_z$, and $G$ has the same unit of acceleration of gravity (G). Solid line and circles, subject A; long dashed line and squares, subject B; short dashed line and triangles, subject C; dotted line and diamonds, subject D.](image)
cause of the diaphragmatic contribution and a minimum contribution of the rib cage (3).

To reconsider the above question, taking into account the whole VC volume range, it may be useful to start evaluating the volume-pressure curve of the thorax in a weightless condition, as distortions due to gravity dependence are abolished, and, therefore, the mechanical properties of the thorax only reflect its elastic features, including the behavior of its two main components (rib cage and abdomen).

At 0 Gz (Fig. 4), the compliance of the thorax is relatively high and fairly constant in the volume range >20% VC, whereas it decreases sharply below this volume and approaching RV, with the resting volume being ~47% VC (Fig. 4, Table 3). The increase in compliance for volumes >20% VC and the observation that the chest wall elastic recoil is low in this volume range suggest that ligaments, tendons, and joints are essentially unstretched at 0 Gz. Therefore, gravity represents the extrinsic factor causing the deformation of the chest wall. However, an intrinsic factor should also be considered, as chest wall compliance was found to vary about threefold at 0 Gz among the subjects.

The elastic behavior of the thorax changes markedly according to whether the gravity vector is in the cranio-caudal (Gz) or in the anteroposterior direction. The changes in position of the volume-pressure curves of the thorax on increasing Gz may be interpreted as the result of the mechanical interaction between the elastic and the gravitational forces acting on the chest (13, 14, 21). The Gz renders the chest wall much stiffer, as mean compliance decreases when Gz grows from 0 to 1 and 1.8 (Table 3, Fig. 4).

For lung volume <30% VC, gravity has an inspiratory effect, but this effect is much larger going from 0 to 1 Gz than from 1 to 1.8 Gz. For volume from 30 to 70% VC, the effect is inspiratory going from 0 to 1 Gz, but expiratory from 1 to 1.8 Gz (the volume-pressure curve at 1.8 Gz shifts to the right of the curve at 1 Gz). For volume greater than ~70% VC, gravity always has an expiratory effect. Loring et al. (14) also found an inspiratory effect on increasing Gz around end-expiratory volume and interpreted the finding by a linear model. The assumption of the linearity actually contrasts with the present results when the whole range of VC is considered. In fact, the chest wall does not behave as a linear system, because the effect of gravity on chest wall depends on both volume and magnitude of Gz.

In the supine posture, similar to the 0 Gz, the volume-pressure curve of the thorax is shifted to the right compared with 1 Gz. The weight of the abdomen in the supine posture caused, on the average, a marked decrease in the resting volume of the thorax from 64 to 31% VC, ~1,650 ml (Fig. 4), and, therefore, exerts an expiratory effect.

The analysis above only applies to the average curves shown in Fig. 4. In fact, if the individual volume-pressure curves are considered, the conclusions could somehow differ because of the variability in the position and the slope of the volume-pressure curves.

The effect of changing Gz on chest wall compliance. The individual differences in compliance at 0 Gz are essentially reduced when subjects are exposed to 1 Gz, where the mean compliance value, 0.19 l/cmH2O, is close to the average reported value of 0.2 l/cmH2O (2). It appears that, despite individual differences in the elastic properties of the thorax in 0 Gz, adding the gravity factor generates a loading that, already at 1 Gz, brings the system close to its minimum compliance.

The gravity dependence of compliance shows a nonlinear behavior in all of the subjects (Fig. 5). However, the analysis of individual curves suggests that the more rigid the chest at 0 Gz, the more its mechanical behavior tends to approach linearity, reducing its sensitivity to the mechanical loading. In fact, coefficient \( yG \) decreases with decreasing \( C_0 \) value, as suggested by the shape of the relationships in Fig. 5.

The nonlinear behavior of the chest wall compliance may result from alinearity of the rib cage and/or abdominal compliance or from the interaction between the two components.

Data from Estenne et al. (7) indicate that abdominal compliance in the tidal volume range does not decrease linearly between 0 and 3 Gz, as its decrease is larger going from 0 to 1 Gz than from 1 to 3 Gz. These data appear qualitatively in line with the present results, although we have no indications about rib cage and abdominal compliance at high and low lung volumes, where the gravity dependence exerts opposite effects on the chest wall. Therefore, the alinearity of the chest wall on changing lung volume and Gz might require a complex modeling of the abdominal-rib cage interaction.

Finally, one may recall that the decrease in compliance of the chest wall going from 1 to 1.8 Gz (~20%), reflecting the increased load applied to the chest wall in the cranio-caudal direction, is mechanically similar to that of the obese subjects, although, in the latter, the decrease in compliance was found to be much larger, ~65% (15).

Static lung volumes. In the supine posture, we found a significant decrease in VC, similar to that reported by Elliott et al. (4) and Agostoni and Mead (1). Some variability, however, occurs relative to changes in static lung volumes concerning 0 Gz. We found a slight (not significant) decrease in VC, a result shared by other investigators (4, 16), and no change in RV, similar to the results found by Kays et al. (12) and Paiva et al. (16) and unlike the results reported by Elliott et al. (4). In our case, the position of the legs in the supine posture could have influenced lung volumes, although data from 30° head-up tilt were not significantly different from supine. We ignore, of course, a possible consequence of mechanical properties of the chest by keeping the legs up rather than horizontally extended.

We do not report data on functional residual capacity that corresponds to the resting point of the respiratory system. Its mechanical definition is the volume at which lung and chest wall exert a pressure at the pleural level (Ppl), equal in modulus and opposite in sign. It appears impossible at present, given the indi-
individual differences in volume-pressure curves of the chest, to define mechanically the changes in functional residual capacity without considering also possible differences in volume-pressure curves of the lungs.

**Comparison with ground-based simulation of 0 Gz.**

The supine posture and water immersion of seated subjects up to the xiphoid have been proposed as models to approximate the effect of 0 Gz on chest wall mechanics (1, 11).

In the supine posture, the rib cage-abdomen mechanical coupling results in a marked decrease in resting volume (similar to what has been previously observed), an effect that can be attributed to the weight of the abdomen contents that displaces the diaphragm cranially.

Estenne et al. (8), using a completely different approach, found no change in compliance going from head up to supine, whereas we actually found an \(2\) to 6.8 times increase. These authors also found that, in the supine posture relative to head up, the rib cage becomes stiffer, whereas the diaphragm-abdominal compartment becomes more distensible. In view of this finding, we might tentatively interpret the increase in stiffness of the thorax, comparing 0 Gz to the supine posture (Fig. 4), as due to the distortion of the rib cage caused by gravity acting in the anteroposterior direction.

In 0 Gz, we found that the resting volume of the chest wall was at 47\% VC, a value that compares reasonably well with that observed in water immersion up to the xiphoid by Agostoni and Mead (1), \(50\%\) VC, and by Hong et al. (11), \(35\%\) VC. However, the compliance of the thorax at 0 Gz was found to be from 2.5 to 6 times higher than during water immersion (1, 11). This difference suggests that, although water immersion counteracts the weight of the abdomen, it does not remove the weight of the rib cage and possibly causes a distortion of the lower thorax.

**APPENDIX**

A statistical approach was adopted to identify the shape of the static volume-pressure curve of the thorax from the original scattered point resulting from the relaxed isovolume maneuvers.

The individual volume-pressure relationship of the chest varied substantially in shape and position (see Fig. 3) in the various conditions studied. The aim of this analysis was not to determine a mathematical model of the volume-pressure characteristic of the thorax, but to confirm, through a numerical and statistical method, the rightness of the curve identified.

From the equations list available, two of them proved useful in finding the best fit for each relationship, the “Boltzmann sigmoidal” and the “double-exponential decay.”

The Boltzmann sigmoidal equation was of the type

\[
y = y_{\text{min}} + \frac{(y_{\text{max}} - y_{\text{min}})}{1 + e^{-x/x_0}}
\]

where \(y_{\text{min}}\) and \(y_{\text{max}}\) represent the minimum and the maximum value of the sigmoidal curve, respectively; \(x_0\) is the pressure corresponding to the midvalue between \(y_{\text{min}}\) and \(y_{\text{max}}\); and \(S\) is the slope of the relationship in the midrange of pressure.

The double-exponential decay equation was of the type

\[
y = a_1 \cdot e^{-k_1 \cdot x} + a_2 \cdot e^{-k_2 \cdot x} + y_c
\]

where \(a_1\) and \(a_2\) are the amplitude of the exponential, \(k_1\) and \(k_2\) are the inverse of the time constant of the exponential, and \(y_c\) is the asymptotic value.

We used a recursive curve-fitting algorithm, based on the Marquartd method, to calculate parameters of the equation chosen, thus minimizing the “sum of the squares” of the data.

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**Table 4. Details of equations used to fit the individual chest wall volume-pressure curves**

<table>
<thead>
<tr>
<th>Subject</th>
<th>Type</th>
<th>1 Gz</th>
<th>Type</th>
<th>1.8 Gz</th>
<th>Type</th>
<th>0 Gz</th>
<th>Type</th>
<th>Supine</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>equation</td>
<td>(r^2)</td>
<td>equation</td>
<td>(r^2)</td>
<td>equation</td>
<td>(r^2)</td>
<td>equation</td>
<td>(r^2)</td>
</tr>
<tr>
<td>A</td>
<td>Sigmoidal</td>
<td>0.95</td>
<td>Sigmoidal</td>
<td>0.92</td>
<td>Exponential</td>
<td>0.91</td>
<td>Exponential</td>
<td>0.80</td>
</tr>
<tr>
<td>B</td>
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<td>0.90</td>
<td>Exponential</td>
<td>0.85</td>
<td>Exponential</td>
<td>0.80</td>
<td>Exponential</td>
<td>0.99</td>
</tr>
<tr>
<td>C</td>
<td>Sigmoidal</td>
<td>0.89</td>
<td>Exponential</td>
<td>0.87</td>
<td>Sigmoidal</td>
<td>0.80</td>
<td>Sigmoidal</td>
<td>0.80</td>
</tr>
<tr>
<td>D</td>
<td>Exponential</td>
<td>0.81</td>
<td>Exponential</td>
<td>0.89</td>
<td>Sigmoidal</td>
<td>0.94</td>
<td>Exponential</td>
<td>0.83</td>
</tr>
</tbody>
</table>

---

**Fig. 6.** A: example of good and homogeneous distribution of relaxation data points over the VC range at 1 Gz. Fitting is described by a Boltzmann sigmoidal equation. B: a case of relative paucity of relaxation data points in the mid-to-low range of lung volumes at 0 Gz. Fitting was obtained by a double-exponential equation.
vertical distances between the points and the curve. An $F$ test was adopted to compare results from the two equations and to determine the best fit equation.

Table 4 reports, for each subject at 1, 1.8, and 0 Gs, and supine posture, the best equations resulting from the analysis and the regression coefficient.

As evident from Table 4, either the double-exponential decay equation or the Boltzmann sigmoidal was able to model the volume-pressure relationship of all subject in all conditions. Despite the good results obtained by the curve-fitting procedure, it was not possible to identify a unique model for the volume-pressure static curve of the thorax. The difficulty arises from the complex matching between the change in shape and position of the relationships, as well as from the variability of the database. This is well explained by looking at data shown in Fig. 6. In Fig. 6A, it is clear that the data points are homogeneously distributed over the VC range, and the fitting was described by a Boltzmann sigmoidal. In Fig. 6B, a more difficult case is presented, where an obviously different shape is accompanied by a relative paucity of experimental observations in the mid-to-low range of lung volumes. One may further observe that, in this case, it appears experimentally difficult to obtain reliable relaxation data in this volume range because of the sharp change from high to low compliance. In this case, the best fit was a double exponential.

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