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Simple and easy to build mechanical ventilator for pandemic health crises in resource-limited areas

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Background: The COVID-19 pandemic exposed critical shortages of mechanical ventilators, particularly in low-resource settings. Disruptions in global supply chains and dependence on specialized components highlighted the need for scalable, locally manufacturing alternatives for emergency respiratory support.
Aim: To describe and evaluate a simplified mechanical ventilator with minimal supply-chain dependence, assembled from widely available automotive and simple hardware components, and intended as a last-resort solution.
Methods: The ventilator is based on a reciprocating air pump driven by an automotive windshield wiper motor coupled to parallel shaft bellows and readily assembled passive membrane valves, only requiring materials available from standard hardware retailers, minimal tools, and basic manual skills. Ventilator performance was assessed through bench testing using a patient model simulating severe lung disease in an adult (R = 20 $cmH_2O{\cdot}s/L$, C = 15 mL/cmH_2O) and pediatric (R = 50 $cmH_2O{\cdot}s/L$, C = 10 mL/cmH_2O) patients. Realistic proof of concept was performed in four mechanically ventilated 50-kg healthy pigs.
Results: According to the bench test, the device delivered tidal volumes up to 600 mL and respiratory rates up to 45 breaths/min with PEEP up to 10 cmH_2O, covering most pediatric and adult ventilation ranges. The *in vivo* proof-of-concept showed that the ventilator maintained arterial blood gases within the targeted range. Technical details for ventilator construction are provided in an open-source video tutorial.
Discussion: Notwithstanding its limitations, this low-cost ventilator demonstrated adequate performance as a prototype. Although not a substitute for commercial intensive care ventilators, its simplicity, autonomy, and low dependence from fragile supply chains provides an approach towards life-saving options in resource-constrained emergency scenarios.

KEYWORDS
emergency, last-resort therapy, low-resource, mechanical ventilation, open-source, respiratory pandemics

1 Introduction

The COVID-19 pandemic exposed the world to an unprecedented health crisis, resulting in a high mortality (1, 2) primarily driven by respiratory failure (3). The scale and severity of the disease fostered an extraordinary demand for mechanical ventilators that exceeded the capacity of even the most advanced healthcare systems (4, 5). This situation necessitated urgent modifications to medical device regulations and exceptional procedures to facilitate the provision of mechanical ventilation and enable survival among all severely affected patients (6–10). The COVID-19 crisis has highlighted that the world remains at constant risk of infectious pandemics –whether due to new influenza variants, emerging coronaviruses or other (11, 12)– that primarily affect the lungs and require respiratory intensive care. The relevant question is no longer whether another such pandemic will occur, but when it will emerge and spread globally, underscoring the urgent need for appropriate preparedness strategies (11).

The experience of the COVID-19 pandemic prompted high-income countries to strengthen preparedness strategies and highlighted the importance of maintaining strategic stocks of ventilators and associated consumables, developing resilient supply chains, and establishing contingency plans for rapidly scaling up manufacturing (6, 13). Unfortunately, the issue of preparedness remains virtually unresolved in low- and middle-income countries, where baseline shortages in critical care resources persist even in the absence of a sudden emergency (14). During a respiratory pandemic surge, low-resource regions can scarcely rely on large-scale provision of ventilators from high-income countries. Even in the absence of financial constraints, health systems in developed nations are themselves likely to be overwhelmed, limiting their capacity to supply external support. Furthermore, the feasibility of locally manufacturing ventilators, based on conventional technology or simplified innovations proposed during the COVID-19 pandemic (15, 16), remains limited in the most deprived settings. Indeed, such initiatives typically require a baseline level of technological infrastructure and specialized expertise that is often unavailable in resource-constrained regions. Moreover, global health crises profoundly disrupt international supply chains for critical technical components, as evidenced during the COVID-19 pandemic (17), thereby restricting access to essential parts required for ventilator assembly.

Under these circumstances, a more realistic strategy for expanding urgent access to mechanical ventilation in low-resource settings during a pandemic would involve the development of a scalable and rapidly deployable mechanical ventilator platform. Such a system should be designed to allow local device assembly by available personnel and constructed from materials that are readily accessible across diverse settings. Importantly, this approach should avoid dependence on fragile global supply chains for specialized technological and medical components, thereby enhancing resilience during periods of global disruption. Accordingly, our aim was to describe an approach providing a potential solution that enables the simple construction of a performant ventilator using widely available components. By returning to core physiological and engineering concepts, the proposed ventilator minimizes technological complexity while preserving the essential functions required to support severely affected patients. Following an open-source publication approach (18, 19), the device's construction process is described in detail to facilitate reproducibility and local implementation. The ventilator performance was evaluated through bench testing, followed by preclinical assessment in a healthy porcine model. Remarkably, the ventilator presented herein is not intended to replace commercially available intensive care equipment. Rather, it is conceived as a potential last-resort alternative designed to provide lifesaving ventilatory support in contexts where no other alternatives are available. In settings where access to mechanical ventilation is otherwise unattainable, such a solution may substantially reduce preventable mortality during respiratory pandemics.

2 Methods

2.1 Ventilator description

The proposed volume-controlled ventilator is based on a classical mechanical configuration using a reciprocating air pump and inspiratory and expiratory valves (Figure 1). The originality of the device presented herein lies in that it requires only common and particularly cheap automotive parts and other simple materials available from standard hardware retailers, minimal tools, and basic manual skills. The ventilator is briefly described below, and full details and construction instructions are provided by a video tutorial in the Supplementary materials.

The cycling air pump (Figure 1) is actuated by a standard car windshield wiper motor (Figure 2) which directly provides reciprocating linear displacement and typically operates at 40–60 cycles per minute when powered by a 12-V car battery. A simple DC motor speed regulator, commonly available as an inexpensive device for controlling automotive fan speed, is used to adjust the cycling frequency. This allows the respiratory rate to be set within the range required for mechanical ventilation in both pediatric and adult patients (12–45 breaths/min).

The reciprocating motor is used to drive shaft bellows (Figure 2) which are widely used in most car models to protect boots that surround drive shaft joints. To achieve the tidal volume (V_T) required for adult ventilation, two such bellows, sealed at both ends with caps, are mounted in parallel and actuate simultaneously using a shared

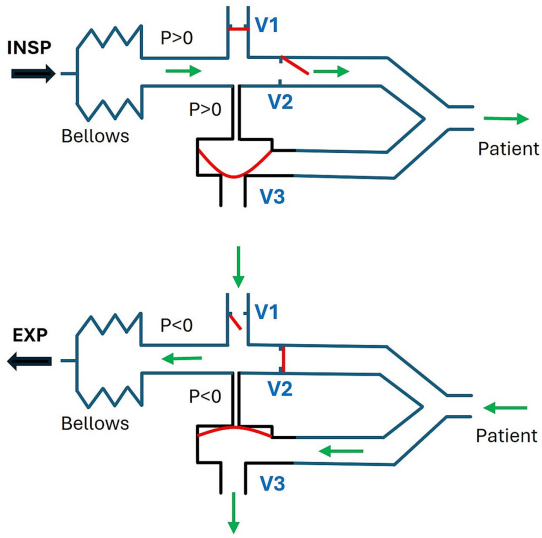


FIGURE 1
Ventilator diagram showing the open or closed status of valves V1, V2, and V3 during inspiration (INSP) and expiration (EXP). Flaps in valves V1 and V2 and membrane in valve V3 are shown in red. Black arrows indicate bellows movement. Green arrows indicate air flow.

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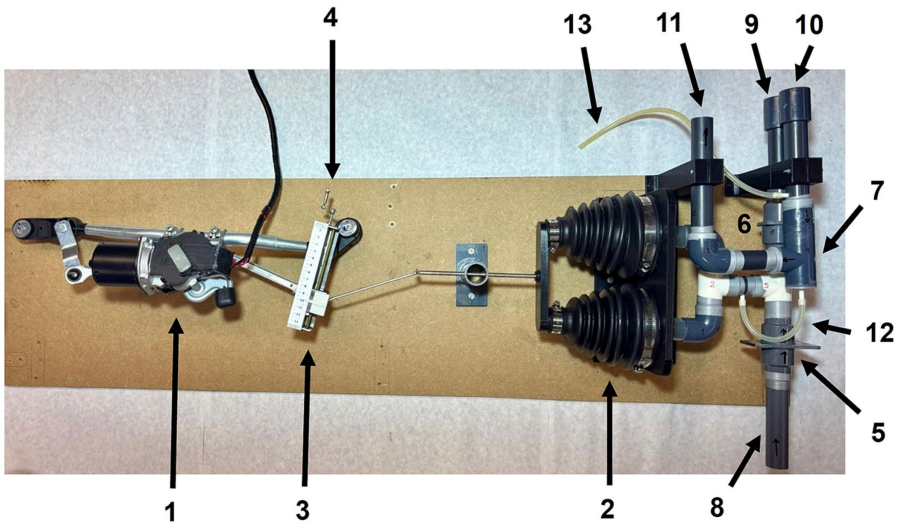


FIGURE 2
Connection of the ventilator basic components: Windshield wiper motor (1), drive shaft bellows (2), variable-length adjustment mechanism (3) including a crank for adjusting bellows displacement (4), flap valves in the inspiratory lines (5, 6), expiratory valve (7), room/O2 air inlet (8), connector for inspiratory (9) and expiratory (10) tubing to the patient's airway, expiratory outlet (11) for connection to a PEEP bottle, tube to drive the expiratory valve by pump pressure (12), tube for sensing airway pressure with a U-tube manometer (13). Detailed construction instructions are provided by a video tutorial in Supplementary materials.

mechanical linkage. A rigid connector links both bellows to the actuation mechanism (Figure 2), ensuring balanced force distribution during compression and synchronized movement. The windshield wiper motor is mechanically connected to the moving end of the bellows via a rod that produces linear back-and-forth to drive the air pump (Figure 2). The connecting rod incorporates a variable-length adjustment mechanism that allows manual modification of the bellows displacement amplitude (and thus V_T) by means of a crank, without interrupting ventilation (Figure 2). This adjustment system is constructed using commonly available hardware materials, including a threaded rod and nuts.

The ventilator valves are passively operated by the reciprocating pump and therefore do not require active control elements (Figure 1): V1 and V2 are flap (i.e., differential pressure-driven) valves, and V3 is based on a flexible membrane passively controlled by the bellows compartment pressure (Figure 1). These valves are manufactured using simple and widely available materials. V1 and V2 membranes are cut from plastic folder sheets, while the flexible membrane for V3 is cut from latex or nitrile clinical gloves. Both valve types are assembled using PVC pipes and fittings (Figure 2). The air inlet valve (V1) allows direct intake of room or oxygen-enriched air (Figure 2). Positive end-expiratory pressure (PEEP) is set using the classical method of immersing the expiratory outlet tube into a water column at a height corresponding to the target PEEP (Figure 3).

A water-filled transparent U-tube (5-mm ID, 45-cm arm length) connected to the inspiratory outlet allows continuous monitoring of airway pressure, particularly peak inspiratory pressure, which is a key parameter for quantifying the magnitude of mechanical ventilation delivered to the patient (Figure 3). An adjustable resistance (constructed from the roller clamp of a conventional intravenous infusion set) is placed between the two arms of the U-tube to optimize the dynamic response of the manometer at ventilation frequencies (Figure 3). Importantly, the U-tube manometer also functions as a safety valve, preventing dangerous lung overpressure in the event of expiratory valve failure resulting in persistent closure during expiration.

2.2 Technical performance: bench test

Ventilator bench testing was performed using a physical model that mimics the mechanical properties of the respiratory system (SmartLung Adult, IMT Analytics, Switzerland), consisting of adjustable resistance (R) and compliance (C) elements (15). These selectable parameters allowed simulation of a wide range of respiratory mechanical conditions representative of different patient populations. Flow and pressure were measured at the entrance of the respiratory model using a pneumotachograph (number 2, Metabo, Epalinges, Switzerland) with a differential pressure transducer (±2cmH_2O; LCVR, Celesco, Canoga Park, CA, USA), and a pressure transducer (±70 cmH_2O; Honeywell 26PC Series, Charlotte, NC, USA), respectively. The electrical current supplied to the ventilator, and therefore its power consumption, was measured. After more than 400 h of continuous ventilation of a patient model, the prototype was tested under conditions simulating healthy lungs and, more importantly, under demanding conditions corresponding to severely diseased adult (R = 20 cmH_2O·s/L, C = 15 mL/cmH_2O) and pediatric (R = 50 cmH_2O·s/L, C = 10 mL/cmH_2O) patients. The respiratory rate was set up to 24 breaths/min and up to 45 breaths/min for adult and pediatric simulations, respectively. PEEP levels of up to 10 cmH_2O were applied. The capacity of the ventilator for gas exchange was tested on the bench by simulating physiological CO_2 production and observing how the ventilator was able to keep normocapnia. To this end, a known continuous flow of CO_2 (ranging 45–350 mL/min) was injected into the compliant component of the patient model and end-tidal CO_2 ($EtCO_2$) was assessed by capnometry (Capnostream 20p, Oridion) at the airway opening (Y-piece of ventilator tubing).

2.3 *In vivo* performance: porcine ventilation

After bench testing, proof of concept for ventilator performance was evaluated in four healthy adult pigs (48–52 kg) following a protocol

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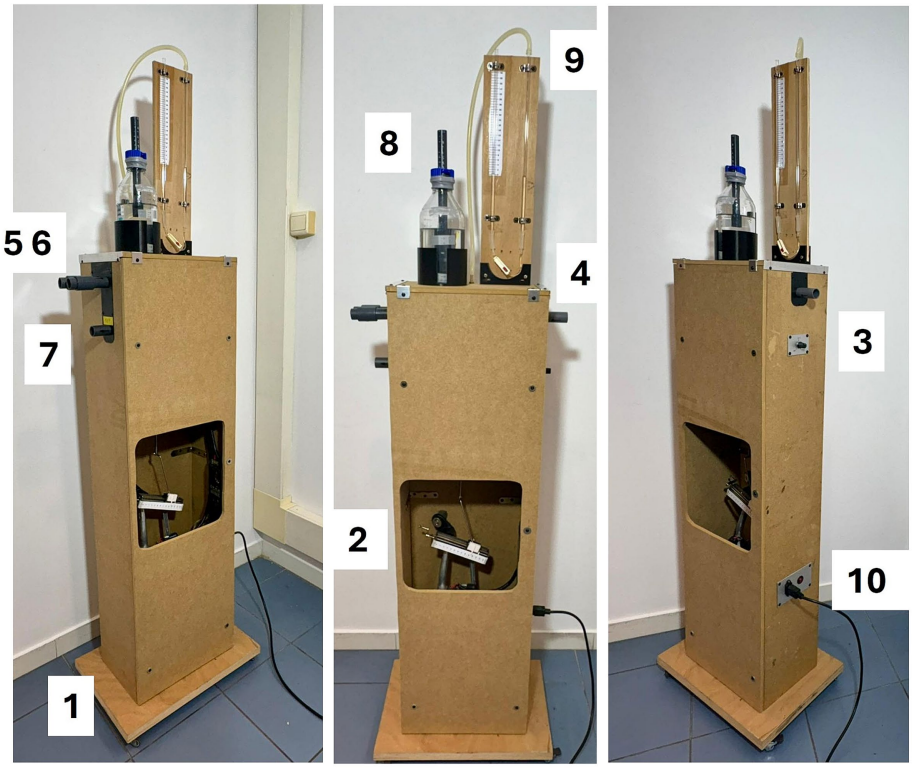


FIGURE 3
Assembled ventilator on a stand-alone platform with wheels (1). The platform in Figure 2 is placed vertically and enclosed in a wooden box: Window for accessing the crank for modifying tidal volume (2), button of the frequency regulator to set breathing rate (3), room/O_2 air inlet (4), connector for inspiratory (5) and expiratory (6) tubing to the patient's airway, expiratory outlet (7) for connection to PEEP bottle (8), U-tube manometer for assessing airway pressure (9), electrical power connection (10) (12-V car battery or conventional 110/220 V AC-line).

approved by the competent ethical board. Commercially available bacterial filters were placed at the air inlet and outlet of the ventilator to be discarded with the tubing after each animal was ventilated. As detailed in the Supplementary materials, the animals were initially instrumented and ventilated using a conventional setting. Once stable physiological conditions were achieved, the commercial ventilator was replaced by the prototype device, with its admission valve V1 connected to an O_2 source through a Venturi valve (from a mask setting) to provide $FiO_2 = 0.4$. From initial settings $V_T = 8$ mL/kg, PEEP = 5 cmH_2O and 20 breath/min, the ventilator was adjusted (mainly breathing rate) as required to maintain arterial PCO_2 within normal values (35–45 mmHg). After 40 min of stable ventilation with the prototype ventilator, physiological data were recorded, and arterial blood samples were collected for gas analysis. The animals were finally euthanized with i.v. infusion of an overdose of sedatives and KCl (1 mEq/kg).

3 Results

3.1 Bench performance

The ventilator performed satisfactorily when used in patient models characterized by markedly increased resistance and reduced compliance. When tested on an adult obstructive–restrictive patient model ($R = 20$ $cmH_2O{\cdot}s/L$, $C = 15$ mL/cmH_2O), the ventilator delivered V_T of up to 600 mL, sufficient to provide protective ventilation (6 mL/kg) for an 85 kg patient. Figure 4 (left) shows representative pressure and flow signals recorded while ventilating this adult model (24 breaths/min, $V_T = 400$ mL) and a high-resistance, low-compliance pediatric model ($R = 50$ $cmH_2O{\cdot}s/L$, $C = 10$ mL/cmH_2O) at a rate of 45 breaths/min with $V_T = 90$ mL (which would correspond to patients with body weight down to <10 kg). The inspiratory flow exhibited a sinusoidal-like pattern quite similar to the physiological inspiratory flow and consistent with the mechanical design of the ventilator air pump. Expiration showed the typical passive flow decay observed in patients. Figure 4 (right) shows the corresponding signals recorded when the same adult and pediatric models were ventilated with PEEP = 10 cmH_2O. In this case, the expiratory portions of both pressure and flow signals displayed slight oscillations caused by bubble formation as PEEP was generated. Electrical power consumption of the ventilator was 7.3 W (PEEP = 0) and 12.0 W (PEEP = 10 cmH_2O) for the adult model, and 18.5 W (PEEP = 0) and 20.1 W (PEEP = 10 cmH_2O) for the pediatric model. By adjusting the ventilation frequency and amplitude within physiological ranges (16–26 breaths/min and 120–500 mL, respectively) the ventilator maintained a normal value of $EtCO_2 = 35$ mmHg for CO_2 productions up to 350 mL/min, thus covering the metabolic activity of pediatric and adult patients.

Performance of the U-tube for pressure monitoring was satisfactory for adult patient ventilation frequencies (up to 25 breath/min) but it decreased for higher pediatric frequencies (Supplementary material). As illustrated by Figure 5, the simple U-tube manometer served as an effective safety valve. After the expiratory outlet of the ventilator was intentionally occluded, when

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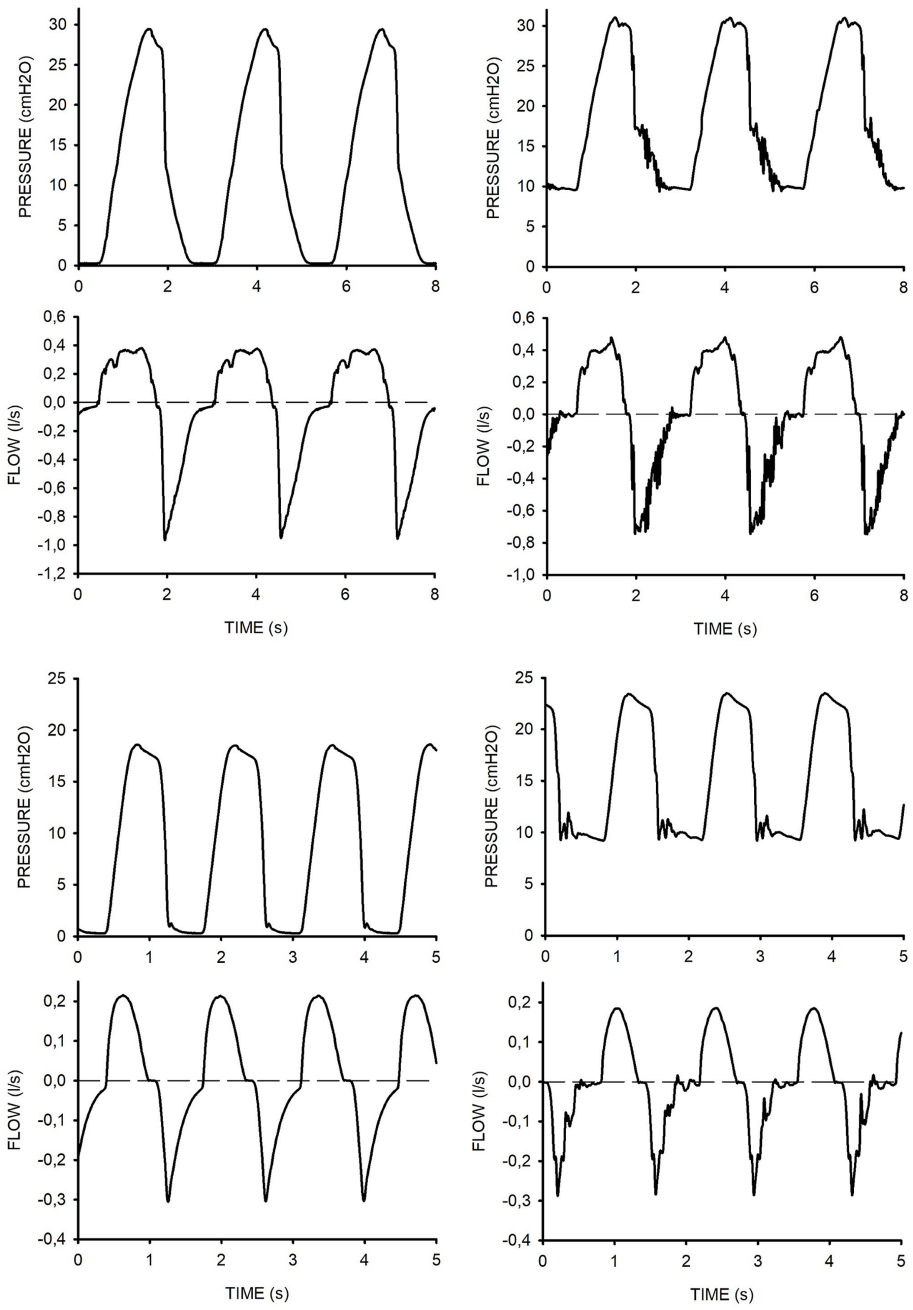


FIGURE 4
Examples of the ventilator performance in the bench test on an adult obstructive-restrictive patient model (top) and on a child patient model (bottom), without (left) and with (right) PEEP = 10 cmH_2O. Pressure and flow (positive during inspiration) were measured at the airway opening of the patient model.

airway pressure reached the level corresponding to the height of the U-tube column, its water was expelled, thereby limiting inspiratory pressure. This mechanism prevented progressive lung overdistension that would otherwise have resulted from continued ventilator cycling while the expiratory valve was abnormally occluded.

3.2 *In vivo* performance: porcine ventilation

As expected from the bench testing results, the ventilator was able to easily and satisfactorily ventilate the 50-kg healthy pigs. With ventilator settings (m ± SD) V_T = 391 ± 7 mL, 19.3 ± 2.1 breath/min and PEEP = 5.5 ± 0.5 cmH_2O, physiological parameters were arterial PCO_2 = 38.3 ± 2.1 mmHg, arterial PO_2 = 188.3 ± 25.0 mmHg (FiO_2 = 0.4), heart rate = 70.7 ± 6.8 bpm, and arterial pressure (max / min) = 137.7 ± 6.8/81.3 ± 3.1 mmHg. Figure 6 shows representative recordings of airway pressure and flow during ventilator setting in one of the pigs (51 kg; V_T = 395 mL), illustrating that the ventilator provided highly repetitive cycles (the coefficient of variation of V_T computed from 10 consecutive cycles was within 1%). PCO_2 was 44 and 35 mmHg for 18 and 23 breaths/min, respectively. Thus, ventilation frequency for this animal was set to 20 breaths/min (PCO_2 = 40 mmHg).

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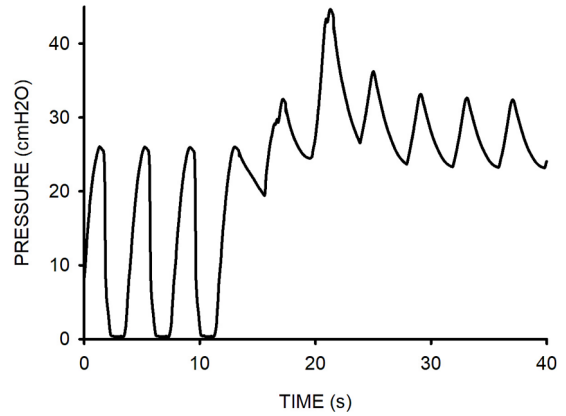


FIGURE 5
Airway pressure recorded during mechanical ventilation when, in the bench test, the expiratory outlet of the ventilator was kept intentionally closed (starting at time ≈12 s). Airway pressure increased to only 44 cm H_2O transiently, avoiding lung over-ventilation.

4 Discussion

4.1 Technical design and implementation of the ventilator

This study was not intended to develop a state-of-the-art ventilator, but rather to design a performant life-saving device easily assembled from components that are widely available across diverse settings, particularly in low-resource environments. To minimize structural and functional complexity, the device is based on early ventilator designs. Accordingly, the device obviates the need for a compressed air source, specialized high-pressure blowers, electronic sensors, commercialized valves and complex control systems. Its technical performance is adequate to deliver ventilatory support across a broad range of respiratory frequencies (up to 45 breaths·min^{-1}) and tidal volumes (up to 600 mL), thereby covering most patients from pediatric to adult body weight categories with PEEP up to 10 cmH_2O (higher PEEP values would require specific testing). In addition to comprehensive bench testing, proof-of-concept of the device was carried out in a porcine model, demonstrating satisfactory functional performance *in vivo*.

The ventilator is constructed around a mechanically driven airway pump coupled with passive membrane-based valves. As the motion-generating element, a standard automotive windshield wiper motor was selected due to its widespread availability and integrated mechanical features. Indeed, this component already incorporates a rotary motor, reduction gear assembly, and mechanical linkage, enabling direct generation of reciprocating motion suitable for bellows actuation in terms of both displacement amplitude and cycling frequency. Ventilatory rate adjustment is achieved through a simple passive regulator, while tidal volume is modulated via a selectable mechanical linkage that modifies stroke amplitude. Notably, windshield wiper motors are engineered for sustained operation under demanding conditions, are powered by a conventional 12-V automotive battery and exhibit low energy consumption. Under high-demand ventilation settings, electrical power measurements in the bench test indicated that the charge stored in a standard 12-V battery car (e.g., 80 A·h) would sustain continuous ventilation for approximately 5.5 and 3.4 days in an adult patient without and with PEEP, respectively. Corresponding autonomy for pediatric ventilation was estimated at 2.2 and 2.0 days, respectively. Any limitation regarding battery autonomy would disappear in case the premise provides conventional AC current since a simple AC (110/220 V) – DC (12 V) converter could be used.

Shaft bellows were selected as the air-displacement element because they constitute a widely available automotive component suitable for ventilator construction. These bellows are typically manufactured from impermeable, flexible, and durable rubber or synthetic polymers engineered to withstand repetitive mechanical deformation under sustained operating conditions. The prototype incorporated two bellows arranged in parallel to achieve the desired volumetric output. However, depending on the specific bellows employed or the target maximum effective tidal volume, the number of parallel units can be increased or reduced to adjust volumetric capacity. Given that for the pressures involved in mechanical ventilation (<40 cmH_2O) the bellows are virtually incompressible, and thus behave as a piston in a cylinder, the crank-displacement scale is useful for adjusting V_T. For a given ventilator (mechanical construct and bellows) the crank-displacement scale can be calibrated in terms of tidal volume by measuring water volume displacement when connecting the ventilator to a bag submerged in water.

The ventilator valves were designed for ease of manufacture, drawing on early ventilatory valve concepts, many of which are currently employed in routine clinical devices such as manual resuscitation bags (20). Flap valves were selected for V1 and V2, as their functional requirement is straightforward: during inspiration –when positive pressure is generated within the bellows– they close and open, respectively; during expiration –when negative pressure develops within the bellows– they open and close, respectively. In contrast, valve V3 requires a different operational principle. This valve must remain closed during inspiration and open during expiration, independently of the pressure gradient between the patient airway and the ventilator outlet during either phase of the respiratory cycle. To fulfill this requirement, V3 was constructed based on early ventilation valve designs (21, 22), consisting of a flexible membrane passively actuated by pressure changes within the bellows chamber (Figure 1). During inspiration, positive pressure within the bellows compartment displaces the V3 membrane to occlude the expiratory limb. Conversely, during expiration, negative pressure within the bellows compartment repositions the membrane, thereby opening the expiratory pathway.

One notable simplification feature of the ventilator is the absence of electronic sensors for monitoring respiratory variables. Instead, the system incorporates a simple U-tube manometer to continuously assess airway pressure (e.g., peak and end-inspiratory values), which is relevant for ventilation monitoring. Importantly, the manometer also enables continuous verification of proper valve functioning, thereby contributing to patient safety. Indeed, if during the inspiratory phase the pressure displayed on the manometer remains close to zero, this finding may indicate one of the following abnormalities: (i) valve V1 remaining open, allowing the air displaced by the bellows to leak into the ambient environment; (ii) valve V2 remaining closed, preventing airflow delivery to the patient; or (iii) valve V3 remaining open, resulting in inspiratory air preferentially escaping to the environment through the expiratory limb. Airway pressure would also remain negligible if V1 is erroneously closed during expiration. In this scenario, ambient air cannot refill the bellows, thereby precluding delivery of the subsequent inspiratory cycle. Interestingly, the manometer may additionally detect malfunctions associated with potential

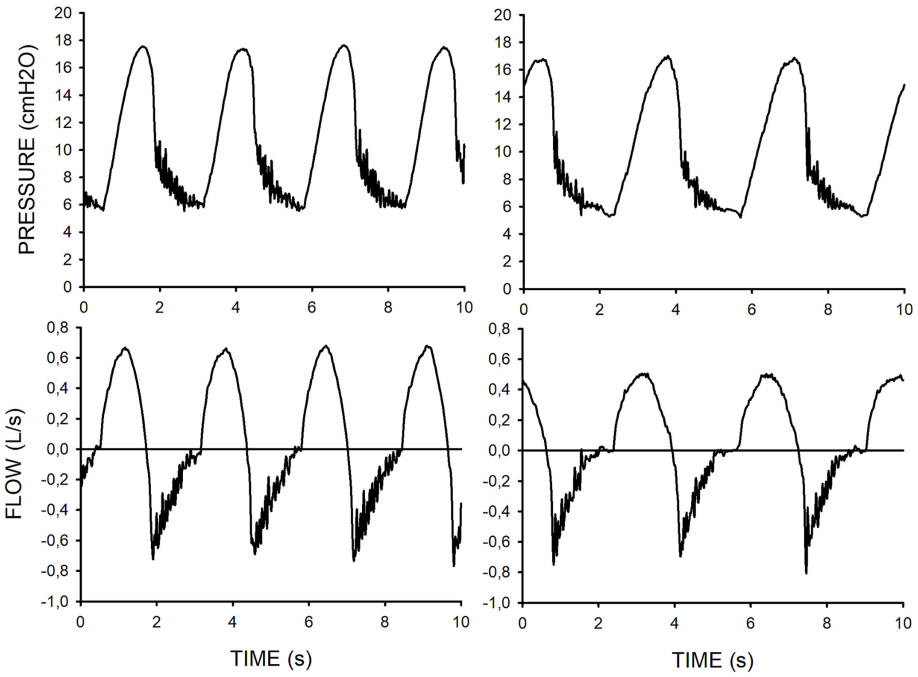


FIGURE 6
Airway pressure and flow (positive during inspiration) recorded when the prototype ventilator was used in a representative pig, with frequencies of 23 breaths/min (left) and 18 breaths/min (right).

rebreathing. For example, abnormal persistence of V2 in the open position during expiration would allow expired gas to enter the bellows. This malfunction would be readily identified by the appearance of negative pressure during the expiratory phase. Finally, as previously illustrated (Figure 5), the manometer also functions as a safety valve in the event of inappropriate closure of V3 during expiration. Moreover, the U-tube manometer can be used to detect potential air leaks throughout the ventilation circuit. Specifically, a decrease in airway pressure when the ventilator is paused at any point during the inspiratory phase (preferably just before the end of inspiration) would indicate the presence of an abnormal air leak.

4.2 Context and potential application of the ventilator

During the COVID-19 pandemic and subsequent years (14, 23) numerous simplified versions of emergency ventilators were proposed (24–29), including designs for multiple patient ventilation (30). Some of them were tested on diseased animal models (24) or patients (28, 29), beyond the healthy porcine model in this study. While these designs represent thoughtful device simplifications with the potential to support safe mechanical ventilation, the majority rely on compressed air source or electronic and medical-grade components sourced from conventional commercial supply chains (e.g., high-pressure blowers, transducers, resuscitation bags, inspiratory/expiratory and PEEP valves). Although such dependence is not problematic under routine circumstances, it constitutes a critical vulnerability during global health emergencies, when supply chains are disrupted (17). The proposed volume-controlled ventilator addresses this limitation through a fundamentally different design strategy. Its fabrication does not depend on specialized technological or medical supply chains. Instead, it utilizes low-cost, locally available mechanical components derived from the automotive sector: a windshield wiper motor, as previously proposed (24, 31), and shaft bellows with easy-to-build valves to replace resuscitation bags. Given the near-ubiquitous global presence of motor vehicles and their associated components, this approach leverages an existing, decentralized industrial infrastructure. Furthermore, the ventilator's open modular architecture enables rapid and straightforward replacement of individual components using spare parts. As a result, this low-cost device offers a potentially scalable and medical supply-resilient alternative for emergency ventilatory support in resource-constrained or crisis settings.

The main limitations of the ventilator design, which stem from its intended simplicity, should be explicitly acknowledged. First, the durability of the automotive parts used as the motor unit and volume generator has not been tested. However, windshield-wiper motors and shaft bellows are originally designed for prolonged operation in vehicles and could therefore be expected to perform satisfactorily as components of emergency ventilators intended for time-limited use. Similarly, the long-term durability of the valves has not been evaluated. Nevertheless, their simple construction and ease of replacement would allow them to be treated as components requiring periodic replacement. Second, by design, the ventilator is constructed using non-medical-grade components, particularly in the parts that come into contact with the inspiratory gas. Third, pressure and volume are assessed only indirectly; therefore, quantitative monitoring of these variables is approximate, and closed-loop control of ventilation is not possible. These limitations were explicitly accepted during the design process and may be considered reasonable for a last-resort solution intended exclusively for situations in which no alternative means of ventilating a patient are available and the patient would otherwise face an almost certain risk of death. Given the specific characteristics of the ventilator and the fact that it is not intended as a commercial

alternative to conventional ventilators, we did not perform a head-to-head comparison with devices equipped with advanced monitoring sensors, cycling controls, and alarm systems. Furthermore, issues such as scalability, deployment logistics, and industrial quality-control processes were not addressed.

A note of caution is warranted. Although the ventilator described herein is relatively straightforward to construct, requires only basic tools, and uses readily available components, this should not be interpreted as implying that individuals without appropriate technical or biomedical expertise can safely build it independently. Safe implementation of the instructions provided in this article and the Supplementary material requires a thorough understanding of the engineering principles underlying mechanical ventilators, as well as the pathophysiological implications of mechanical ventilation. Therefore, even in situations of extreme necessity, the device should be constructed and deployed only under the supervision and responsibility of appropriately qualified engineers and/or physicians.

Historical experience demonstrates that, in large-scale health emergencies, life-sustaining therapies and medical devices are often deployed before comprehensive validation or formal regulatory approval, signaling operational necessity of improvisation when conventional well-established treatments are insufficient or unavailable (32–35). Referring specifically to medical devices in respiratory emergencies, during mid-20th century poliomyelitis epidemics, negative-pressure ventilators and manually operated bag ventilation were implemented at scale despite limited standardization, successfully prioritizing functional respiratory support over technological refinements (36). The COVID-19 pandemic is a more recent example where exceptional actions were taken regarding the usability of mechanical ventilators (5, 6). These precedents demonstrate a consistent general principle in disaster medicine: when mortality risk is immediate and infrastructure is overwhelmed, the threshold for deploying simplified, investigational, or non-conventional technologies shifts toward pragmatic functionality and scalability (37–41). Within this framework, the exceptional use of the ventilator described here, even though it may be previously unregulated and built with non-medical grade components, would align with established vital response paradigms (beneficence, non-maleficence and scientific plausibility) if permission from the local authority in charge of the emergency is granted (40, 41).

5 Conclusion

In the context of a health emergency by respiratory pandemic in a low-resource setting, the proposed ventilator with minimal supply-chain dependence, or a slight variant following the same design strategy, may provide a very last-resort tool to provide respiratory support to patients who otherwise would have no access to this life-saving therapy.

Data availability statement

The original contributions presented in the study are included in the article/Supplementary material, further inquiries can be directed to the corresponding author.

Ethics statement

The animal study was approved by Animal Research Ethics Committee of the University of Barcelona (permission number 694/25), in compliance with the relevant European and National regulations. The study was conducted in accordance with the local legislation and institutional requirements.

Author contributions

RF: Conceptualization, Data curation, Formal analysis, Funding acquisition, Investigation, Methodology, Project administration, Resources, Supervision, Validation, Writing – original draft, Writing – review & editing. RS: Investigation, Methodology, Writing – review & editing. MR-L: Methodology, Writing – review & editing. KK: Investigation, Writing – review & editing. LF-B: Investigation, Writing – review & editing. VO: Investigation, Writing – review & editing. AT: Investigation, Validation, Writing – review & editing. NF: Investigation, Validation, Writing – review & editing. AD-X: Investigation, Validation, Writing – review & editing. DG: Investigation, Validation, Writing – review & editing. JO: Investigation, Methodology, Writing – review & editing.

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Conflict of interest

The author(s) declared that this work was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

The author RF declared that they were an editorial board member of Frontiers, at the time of submission. This had no impact on the peer review process and the final decision.

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Supplementary material

The Supplementary material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fpubh.2026.1855214/full#supplementary-material

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